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Journal of Rehabilitation Research and Development
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The *Journal of Rehabilitation Research and Development,* published quarterly, is a scientific rehabilitation research and development publication in the multidisciplinary field of disability rehabilitation. General priority areas are: Prosthetics and Orthotics; Spinal Cord Injury and Related Neurological Disorders; Communication, Sensory and Cognitive Aids; and Gerontology. The *Journal* receives submissions from sources within the United States and throughout the world.

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GUEST EDITORIAL

The Rehabilitation Icon

Computer icons and jargon have become imbedded in our visual and oral language. To the array of file folders, magnifying glasses, and scissors that head many computer screens, a three-legged stool might be added to symbolize contemporary rehabilitation service. The "pull-down" menu for this icon would feature choices relating to each leg of the stool—research, education, and administration. The icon suggests that rehabilitation has a three-fold support. When all supports are sturdy, rehabilitation service is well sustained. But, if one leg is rickety, the stool is likely to collapse. No leg is more important than any other, each must carry its load. Clicking the computer mouse on this icon would enable one to open the file entitled *Journal of Rehabilitation Research and Development,* for this publication addresses all elements of the rehabilitation model and has done so long before a mouse was anything other than a household pest and icons were found only in monasteries.

Rehabilitation is supported by research. Without ongoing questioning, investigating, and linking new findings with previous experience, our patients would not benefit from new types of prosthetic feet or better means of controlling a painful knee or more effective means of speaking. The scientific articles in the pages of the *Journal of Rehabilitation Research and Development* reflect the ever broadening scope of rehabilitation.

Education is the second leg. Timely dissemination of research results at scientific meetings and through journals of high integrity, such as the *Journal of Rehabilitation Research and Development* enables clinicians distant from the research centers to improve their rehabilitation services to the benefit of the patients they serve.

The third support is administration. Without responsible management of fiscal and human resources, the rehabilitation endeavor collapses. Just as the ongoing excellence of the *Journal of Rehabilitation Research and Development* reflects the high quality of its direction, the entire rehabilitation effort depends on skillful administration.

Joan E. Edelstein, M.A., P.T.
Associate Professor of Clinical Physical Therapy and Director, Program in Physical Therapy, Columbia University, New York, NY

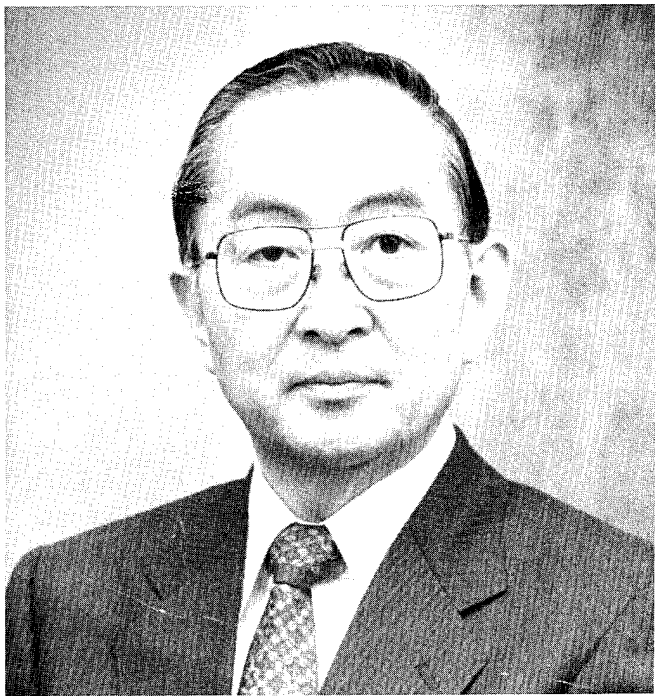
Although the stool icon does not flash on the computer screen, the three-fold construct of rehabilitation is a code which reveals itself in the well-being of our patients.

Joan E. Edelstein, M.A., P.T.

FOREIGN EDITOR'S REPORT

Rehabilitation Research and Development in Physical Medicine in Japan: A Historical and State-of-the-Art Review

The Eighth World Congress of the International Rehabilitation Medicine Association (IRMA VIII) will be held in Kyoto, Japan in August-September, 1997. As one of the Foreign Editors of the *Journal of Rehabilitation Research and Development* from Japan, as well as the chairman of the IRMA VIII Organizing Committee, I take this editorial as a most appropriate opportunity to introduce to American and international readers a short historical and state-of-the-art review of rehabilitation research and development in this Far Eastern country that has recently emerged as one of the world's economic powers, but whose cultural and scientific contribution to the world community still remains relatively unknown.



Satoshi Ueda, M.D.
Professor and Chairman, Department of Rehabilitation Medicine, Teikyo University, Ichihara Hospital, Chiba, Japan

Target of the Service: From Children to Young Adults to the Elderly

The origin of rehabilitation in Japan can be traced to the early 1920s, when Dr. Kenji Takagi, then Professor of Orthopaedic Surgery at the University of Tokyo, urged the medical profession and society in general to support his and his colleagues' devoted work for restoring independence to children with physical disabilities, mainly children with polio. Dr. Takagi considered the traditional orthopaedic approach inadequate for this task and looked for a new way of thinking. Finally, he coined a pair of new Japanese words that expressed his new ideas most appropriately. One was "Shitai-fujiyuu-ji," meaning literally "limb-and-body not-free children"; that was meant to replace the old notions, such as "crippled," "deformed," or "invalid" children, etc., emphasizing the restriction on the freedom of the child as a human being. The other new word was "Ryo-iku," which Dr. Takagi composed taking the first part (ryo) from a Japanese word meaning medical treatment, and the second part (iku) from another word meaning education (general and vocational).

Thus, Dr. Takagi's vision of Ryo-iku (treatment-education) closely resembled the present-day concept of rehabilitation as being a comprehensive, multidisciplinary effort to restore people with disabilities to their fullest possible independence. Dr. Takagi was instrumental in founding the Japanese Association for the "limb-and-body not-free children" in 1925, in establishing the first special school for such children in 1932, and opening the first rehabilitation center for the disabled children in 1942. He was the director of the latter until the end of his life in 1963.

World War II dealt a hard blow to Dr. Takagi's humanitarian endeavors. For example, his center was destroyed by an air raid. However, the wartime experience of rehabilitating veterans with amputation and spinal cord injury in hospitals run by the Agency for Protection of Militarily Wounded (a Japanese counterpart of the U.S. Department of Veterans Affairs) paved the way for the postwar development of rehabilitation services for young adults with disabilities.

In postwar Japan, rehabilitation for children with disabilities flourished anew, while the main cause of

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disability in children shifted rapidly from polio to cerebral palsy in the early 1960s. Thus, the target population of rehabilitation in postwar years was no longer restricted to children but came to include young adults as well.

The rapid aging of the general population, with longer life expectancy resulting from medical progress and the improvement of living conditions, gave rise to the development of geriatric rehabilitation, mainly for stroke, in the 1960s.

Therefore, it is interesting to note that the emphasis of rehabilitation in Japan has shifted historically from children to young adults to the elderly, just as the life stages of a human being.

There has been a maturation process in rehabilitation from the 1970s to the present day, including:

1. the gradual transition of medical rehabilitation services for the elderly away from spa resorts and toward the cities.
2. the development of rehabilitation services in general hospitals allowing an earlier start in the acute stage.
3. expansion of target groups from the traditional persons with amputation, spinal cord injury, cerebral palsy, and stroke to those with neuromuscular disease, traumatic brain injury, lung and cardiac diseases, cancer mental illness, etc.
4. deinstitutionalization and development of community rehabilitation and independent living of people with disabilities.
5. development of coordination and cooperation among medical, educational, vocational, and social rehabilitation professionals.

Professional Organizations and Professional Education

The year 1963 was a very important landmark in the development of rehabilitation-related professions in this country. The Japanese Association of Rehabilitation Medicine (JARM), the most prestigious academic organization in this field, was founded in 1963 by about 100 physicians and surgeons. Now, after 32 years, it has a membership of more than 8,000 MDs with a small number of allied health professionals. In 1980, JARM started a board certification system for physicians whose specialty is in rehabilitation medicine. The number of board-certified specialists is now over 500.

The first rehabilitation service in a university hospital was established in 1963 in the University of Tokyo. Today, there are more than 20 professional chairs in rehabilitation medicine, and medical education in rehabilitation medicine is included, more or less, in all of the 80 medical schools in Japan.

The first school for physical and occupational therapists was opened in 1963 in Kiyose, Tokyo, with a 3-year diploma course. The first national examination for both the professions was given in 1966, and the Japanese Association of Physical Therapists (JAPT) and the Japanese Association of Occupational Therapists (JAOT) were founded in the same year. Now, after 29 years, JAPT has a membership of more than 10,000 and JAOT of about 6,000. The first junior college education (a 3-year course) in physical and occupational therapy was started in Kanazawa University in 1979. Finally, the first 4-year college course in physical and occupational therapy was started in Hiroshima University in 1992. Currently, there are more than 70 physical therapy schools and more than 50 occupational therapy schools, which means that Japan has the second largest number of physical and occupational schools in the world.

Trends in Rehabilitation Research and Development

Research in rehabilitation medicine and engineering is conducted in medical schools, in large recreation centers sponsored by either national or prefectural governments, in research institutes, in large rehabilitation hospitals, etc. The most favored topics for research in rehabilitation medicine are stroke, spinal cord injury, amputation, cerebral palsy, rheumatoid arthritis, muscular dystrophy, and traumatic brain injury. Those in rehabilitation engineering are prosthetics, orthotics, communication aids, architectural barriers, and designs for people with disabilities.

The annual congress of the Japanese Association of Rehabilitation Medicine offers special lectures, symposia, panels, and seminars, and generates more than 400 scientific papers, all related to the above-mentioned topics. Both national associations of physical and occupational therapists (JAPT and JAOT) hold their annual congresses with comparably large programs.

Also, there is an annual Biomechanism Symposium on basic and applied research and development in rehabilitation engineering and related fields.

There are three major Japanese monthly journals on rehabilitation medicine, two monthly and one bimonthly journals on physical therapy, and one monthly and one quarterly journal on occupational therapy. Currently, there is no journal on rehabilitation engineering except for the annual proceedings of the aforementioned Biomechanism Symposium.

Toward the Twenty-first Century

Rehabilitation research and development in Japan, having come of age, has much to contribute to the body of knowledge and experience in the world community of professionals working together for the benefit of the people with disabilities. We must look forward to the twenty-first century that is within our reach. Our task now is to break through the communication/information gap that is due mainly to language barriers.

Satoshi Ueda, M.D.

FROM THE EDITOR

Seishi Sawamura, M.D., one of the *Journal's* Foreign Editors, is newly elected as President of the International Society for Prosthetics and Orthotics (ISPO) and its Executive Board, as well as being chairman of the Japanese Society of ISPO. He is currently the international consultant for all of Southeast Asia. As such, he provides information regarding prosthetics, orthotics, and rehabilitation activities.

Dr. Sawamura and Eiji Tazawa, who has been appointed as the second international consultant to Southeast Asia, are both having discussions with representatives of the Indonesian Ministry of Health concerning the possible establishment of an Asian Prosthetics and Orthotics Center in Indonesia. The plan for the center includes a 3-year prosthetic and orthotic education program, a prosthetic and orthotic patient care department, a research center, and a component manufacturing facility. These plans are still in their early stages and it will be necessary to establish a clear picture of the need in order to ensure that the center can continue to function permanently after the Japanese involvement ends.

As Director of the Hyogo Rehabilitation Center in Kobe, Dr. Sawamura was instrumental in the rehabilitation of many of the casualties that resulted from the Kobe earthquake.

Several members of this journal's editorial board are also members of ISPO and were present at the 8th ISPO World Conference in Melbourne, Australia, when Dr. Sawamura was installed as the incoming president by the Executive Board and the outgoing president, now Immediate Past President, Melvin L. Stills, C. P. O., Southwestern Medical Center, Dallas, Texas, USA.

It is our hope that Drs. Ueda and Sawamura will provide us with research and development information in rehabilitation medicine, prosthetics/orthotics, and rehabilitation engineering that may be applicable to our veteran population in the foreseeable future.

Dudley S. Childress, Ph.D., member of the editorial board of this journal since its inception, and VA Rehabilitation Research and Development Service principal research investigator in upper limb prosthetics, was in attendance at this past ISPO World Conference in Melbourne and has a report for you.

Tamara T. Sowell
Editor

Department of Veterans Affairs

8th World Congress of ISPO: A View

Dudley S. Childress, PhD
Northwestern University, Chicago, IL 60611

Venue: The 8th triennial world congress of the International Society for Prosthetics and Orthotics (ISPO) was held in Melbourne, Australia, April 2–7, 1995. It was the first ISPO World Congress in the Southern Hemisphere. Melbourne, the capital of Victoria, is a delightful city on the Yarra River. Its most notable difference from other modern cities is its commitment to trams for public transportation. Trams, as Congress participants found out, are an efficient, clean, and charming means for urban travel along Melbourne's tree-lined streets amid her Victorian architecture. Melbourne is perhaps the most European-like of all the cities in Australia. Like other cities of Australia, it has flora of great beauty in its many parks and the Royal Botanical Garden is an exceptional place in the central city. The Governor's house is located on the Garden grounds and many Congress delegates were received by the Governor of Victoria at a beautiful reception there. As well as being an important business and sports center of Australia, Melbourne is also an important educational and cultural city of the South Pacific. Australia's National Centre for Prosthetics and Orthotics (now a part of La Trobe University) is in Melbourne. Melbourne's former Central Development Unit has expanded and is now the Monash University Rehabilitation Technology Research Unit. This facility has played an important role in prosthetics and orthotics since the end of World War II.

The ISPO event, in collaboration with INTERBOR and coincident with the Society's Silver Jubilee, was held at the World Congress Centre Melbourne, a new convention hall located on the Yarra near the southwestern corner of the city proper. The site proved to be a superb venue for the World Congress. More than a thousand participants from approximately 70 countries took part in the event and they were treated to an excellent program that was supplemented by enjoyable and interesting social occasions.

Organizers: The 8th World Congress was hosted by the Australian National Society of ISPO, with the New Zealand Artificial Limb Board associated as a sponsor. Valma E. Angliss, a well-known and highly regarded physical therapist from Melbourne, was the Secretary General of the Congress. She did an absolutely magnificent job of organizing the Congress. The world community in prosthetics and orthotics can be grateful for her labor and dedication and for the work and dedication of so many other Australians and New Zealanders who made the Congress a great success. The event was in good hands. Of course, many other people played important roles in the meeting, particularly Mel Stills, C.O., of the USA, who had served as President of ISPO for the last 3 years, and Seishi Sawamura, M.D., of Japan, the Melbourne Congress convenor, and the new president of ISPO.

Dudley S. Childress, Ph.D.
Director, Northwestern University, Prosthetics Research Laboratory, Chicago, IL; Director of Prosthetic Research, VA Medical Center, Lakeside, Chicago, IL

Highlights: It is always difficult to select highlights of a meeting. The most obscure paper of some practically unknown investigator, may in fact be the most important contribution of the meeting. Only time can reveal "opals" of this kind among the diggings of a World Congress.

However, obvious highlights can be spotted. Some were:

1. The Symposium on Osseointegration in Prosthetic Orthopaedics. This symposium, presented by a Swedish team from Gothenburg, was an outstanding event, because this was the first time that their extensive and impressive preliminary work on direct skeletal attachment had been presented to the broad prosthetics and orthotics community. From an external prosthesis standpoint, their prosthetic extensions of fingers and thumbs were particularly impressive because direct bone attachment provides a rigid connection to the remaining finger or thumb and thereby enables the replacement prosthesis to have considerable function and to be a direct conduit of sensory information to the body (e.g., through forces and vibration). The reports of direct skeletal attachment of upper and lower limb prostheses were also impressive. Direct skeletal attachment has long been considered desirable in prosthetics. The Swedish investigators have gone a long way in demonstrating the feasibility of this approach. Their previous successes with osseointegration of tooth implants, of skull anchors for support of maxillofacial prostheses, and of directly attached bone conduction hearing aids demonstrates that this work has depth and that it is backed up by years of clinical and scientific studies. Attendees who were also at the 1983 World Congress may have noticed similarities between the excitement of this symposium and the excitement engendered in London at the first presentation of CAD/CAM to the prosthetics community.
2. The recognition of Mr. Sepp Heim as the Knud Jansen Lecturer brought into focus the need for improved prosthetic and orthotic (P&O) services worldwide and the important role that education programs play in bringing about these services. A significant aspect of the Congress concerned delivery of appropriate prostheses and orthoses to people in all countries of the world. Mr. Heim's recognition by ISPO and his lecture highlighted this need. The lecture is published in the most recent publication of *Prosthetics and Orthotics International.* Mr. Heim's work in helping to establish P&O educational programs in Tunisia, Togo, Tanzania, and China demonstrates what a big impact a single, effective person can have.
3. The Melbourne meeting demonstrated the increasing future role that Asian countries are likely to have in the development, manufacture, and marketing of P&O products. Besides products from Japan and Taiwan, products developed in Shanghai, Hong Kong, Beijing, and other locations were shown in the exhibit hall.
4. Congress papers and exhibits showed a continued major trend in prosthetics toward structures that are much more compliant than structures used in the past. A continued trend toward polycentric joints was also evident at the meeting and the combination of these joints with elastic structures to form "bouncy knees" appears to be on the increase. Knees containing computer systems, such as in the "Intelligent Knee," developed in Japan and now being marketed through Endolite, appear to be of increasing importance. An interesting new aspect of the Intelligent Knee is that the prosthetist can "tune" it by remote control, without having to pull down the cosmetic cover to make adjustments.
5. The development of new componentry continues to be ahead of the field's scientific and engineering understanding of how these components function as part of the human-prosthesis or human-orthosis system. Papers and presentations at the meeting illustrated that gait analysis techniques still do not provide as much insight into these human-machine systems as we would hope. An exception appears to be in the surgical management of children with cerebral palsy. The presentations of Dr. James Gage on this topic were well received and were another highlight of the Congress.

Uniqueness: One of the unique aspects of ISPO congresses is the admixture of surgeons, physicians, therapists, prosthetists, orthotists, engineers, technicians, and others who come together creatively at this triennial event. This kind of interaction is particularly evident in the upper limb prosthetics sessions, which are more interesting and vibrant at the World Congress than at more narrowly defined professional meetings of the field. In a similar manner, the Congress is one of the few venues where the members of the surgical community come together with the P&O field. There was a strong surgical component in the program. The interdisciplinary nature of the meeting and the international scope of the participants make the Congress unique.

Awards: In addition to Sepp Heim's recognition as the Knud Jansen lecturer, Össur Christensson of Iceland received the Brian Blatchford Prize for innovative prosthetics designs. Robert J. Gailey of the USA accepted the Forcheimer Prize for himself and associated authors for the paper "CAT/CAM Socket and Quadrilateral Socket—A Comparison of Energy Cost During

Ambulation," which appeared in *Prosthetics and Orthotics International.*

Personal Rumination: International meetings require a lot of effort on the part of the organizers. They take time to attend and they are costly. Nevertheless, for the time being, they seem to be worth the time, effort, and cost expended. Americans learned after World War II that their prosthetic methods and technologies were inferior at that time to those in Europe, and much of the rapid progress in prosthetics in the USA after the war was partially a result of visits by USA teams to key facilities in Europe, particularly Germany. To remain a leader in prosthetics and orthotics services for veterans, VA and others must keep abreast of advances worldwide. For this reason, the *Journal of Rehabilitation Research and Development* needs to remain international in its scope of publications, and VA personnel need to attend world congresses of ISPO, whenever possible. The 9th World Congress of ISPO is to be held in Amsterdam, The Netherlands, June 28–July 3, 1996. It promises to be one of the best meetings ever.

LETTERS TO THE EDITOR

To the Editor:

Re: A survey of marginal wheelchair users JRRD 31(4):297–302

As a manager (and physiotherapist) of a UK London Wheelchair Service I read with interest this paper.

My comment is chiefly that of the need of good assessment and prescription to meet the user's medical, physical, social, psychological, recreational, and educational needs. NHS Wheelchair Services has a remit to meet users' needs and if they are not doing this with a variety of models of wheelchairs (not simply the ministry range, as in the case of 75% in this study), then they are not doing their job.

If they are saying that they have insufficient funding to meet users' needs, then they should be discussing this with the purchasers of their service.

For 59% of users questioned to say that "their wheelchair was inadequate for their requirements," reflects poor assessment and prescription by the professional concerned and is, in my opinion, inexcusable.

Yours faithfully,
Rosalind Ham, FCSP, Msc, Cert. Ed SRP
Superintendent Physiotherapist
Wheelchair Service Manager
St. Andrew's Hospital
City and East London Family and Community Health Services.

ERRATA

In Vol. 32 No. 1, page 57 line 6 word 3 of the legend for Figure 1 in the Special Report by Houston, et al., should read "6.35 mm" rather than "6.35," thus having the sentence read "The wire frame residual limb measurement model is shown at the 10° cross-sectional angular increment and 6.35 mm axial increment resolutions output by most prosthetics CAD/CAM (opto-)electromechanical digitizers."

Page 57 line 6 word 4 of the legend for Figure 2 should also read "6.35 mm" rather than "6.35," thus having that sentence read "The wire frame residual limb measurement model is shown at the 10° cross-sectional angular increment and 6.35 mm axial increment resolutions output by most prosthetics CAD/CAM (opto-)electromechanical digitizers."

Also on page 57 of the same report, the heading on Figure 3a should not have a period after "Modification" and before "Region," thus having the heading read "CAT PTB Socket Design Template Location of Fibular Head Modification Region Vs. BK Amputee Residual Limb Fibular Head Location."

On page 58, the heading on Figure 3b should not have a period after "Modification" and before "Region," thus having the heading read "CAD AK IC Socket Design Template Location of Quadriceps Modification Region Vs. AK Amputee Residual Limb Quadriceps Location."

Finally, on page 58 line 1 column 2, word 8 should read "6.35 mm" rather than "6.35 cm."

Editor
Tamara T. Sowell

Clinical Relevance for the Veteran

SUMMARY OF SCIENTIFIC/TECHNICAL PAPERS IN THIS ISSUE

Energy Expenditure During Ambulation in Dysvascular and Traumatic Below-Knee Amputees: A Comparison of Five Prosthetic Feet.

Leslie Torburn, MS, PT; Christopher M. Powers, MS, PT; Robert Guiterrez, MD; Jacquelin Perry, MD *(p.111)*

Purpose of the Work. Dynamic Elastic Response (DER) prosthetic feet are reported to help persons with amputation walk by reducing the amount of energy required. The purpose of this study was to determine if there were any differences in energy expenditure between the DER feet and the traditional SACH foot. **Subjects/Procedures.** Nine traumatic and seven dysvascular amputees participated in this study. Each subject underwent energy cost testing while wearing five different prosthetic feet (SACH, Flex-Foot, Seattle, Quantum, and Carbon Copy II). A 1-month accommodation period was given for each foot prior to testing. **Results.** There were no differences in energy cost or rate of energy expenditure between any of the prosthetic feet tested. The traumatic amputees had a similar energy cost compared to the dysvascular amputees; however, the rate of energy expenditure was higher in the amputee group. **Relevance to Veteran Population.** This study shows that there is no apparent energy cost savings with the DER feet compared to the traditional SACH foot. This implies that the DER feet do not address the loading of the amputated limb which has been shown through electromyography (EMG) to be the greatest period of muscular demand.

Christopher M. Powers, MS, PT

Gait Initiation of Persons With Below-Knee Amputation: The Characterization and Comparison of Force Profiles.

Stephen A. Rossi, BSME; William Doyle, MS; Harry B. Skinner, MD, PhD *(p. 120)*

Purpose of the Work. Persons with below-knee amputation compensate for lost ankle function in their performance of activities of daily living, especially for the complex task of gait initiation. The goals of this study were to characterize the forces involved in gait initiation and identify the effects of prosthetic alignment. **Subjects/Procedures.** Seven males with below-knee amputation were recruited through the VA Medical Center, San Francisco. These subjects underwent gait initiation trials while varying prosthetic alignment. Force data were collected for each initiation. **Results.** Significant asymmetries were found in the force profiles of the amputated and non-amputated limbs; forces were consistently higher for the prosthetic limb. Changes in prosthesis alignment proved not to have significant effects on generalized force parameters. **Relevance to Veteran Population.** With the large Veteran amputee population, our results indicate that efforts should be directed toward persons with amputation in the early stages of gait training before asymmetries become habits.

Harry B. Skinner, MD, PhD

Prediction of Metabolic Energy Expenditure During Gait From Mechanical Energy of the Limb: A Preliminary Study.

Scott A. Foerster, MS; Anita M. Bagley, PhD; C. Dan Mote, Jr., PhD; Harry B. Skinner, MD, PhD *(p. 128)*

Purpose of the Work. Energy consumption of persons with transfemoral amputation during walking is an important determinant of the maximum distance and speed they can achieve. It would be convenient if relatively easily measured variables, such as the mechanical motions of a prosthesis during gait, gave reliable information about more difficult measurements, such as total metabolic energy consumption. This study was designed to determine if measurement of the energy of prosthesis motion was a reliable indicator of amputee energy consumption. **Subjects/Procedures.** Metabolic oxygen consumption during gait was measured for four transfemoral amputees. And, the work done by their prostheses was determined by measuring the motions and masses of the prosthesis. **Results.** Mechanical work done by the subjects' prostheses was not a good indicator of metabolic oxygen consumption for these amputees. **Relevance to Veteran Population.** While larger numbers of subjects may demonstrate a relation between these variables, the relation is not good enough to be used in managing the rehabilitation of persons with transfemoral amputation.

Harry B. Skinner, MD, PhD

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Effects of Thigh Soft-Tissue Stiffness on the Control of Anterior Tibial Displacement by Functional Knee Orthoses.

Stephen H. Liu, MD; J. Michael Kabo, PhD; Aaron Daluiski, MD *(p. 135)*

Purpose. To examine the effects of thigh stiffness (thigh muscle bulk) on the ability of commercial custom functional knee orthoses to control anterior tibial displacement. **Subjects/ Procedures.** Stiffness of the human thigh was measured in 10 competitive, 10 recreational, and 10 sedentary subjects. Three soft tissue analogs were reconstructed with soft and ridged foam layers with similar soft-tissue stiffness as the subjects. The soft-tissue analog was applied to the surrogate knee for testing of the functional knee orthosis. During testing, one of the soft-tissue analogs was randomly selected and attached to the surrogate knee femur and a randomly selected orthosis was applied to the surrogate limb. Anterior tibial forces ranged from 25 to 200 N applied at 25 N increments, and the anterior tibial displacement was recorded following application of each force. Using the sequence, each orthosis was tested 11 times for each soft tissue analog. The anterior tibial displacement at each applied force was recorded. The entire sequence of 11 trials was repeated for the four different custom knee orthoses. **Results.** All three soft-tissue analogs on the control of anterior tibial displacement were variable for the different braces; however, in general, the least stiff soft-tissue analog resulted in greater anterior tibial displacement, especially at high forces. Townsend and DonJoy custom orthoses showed a decrease in average tibial displacement with increasing stiffness at all forces measured. **Relevance to Veteran Population.** Anterior cruciate ligament injuries are common among the young active veteran population. Custom knee orthoses have been used to control knee instability. However, no studies have looked at the effect of improved quadriceps muscle strength and bulk with the use of functional knee orthoses. In our study, we have demonstrated that increased thigh soft-tissue stiffness will facilitate a functional knee orthosis in reducing anterior tibial displacement. Thus, in order to employ the functional knee orthosis most effectively, rehabilitation to improve quadriceps strength and bulk will be important not only for older persons with less muscle mass, but also for persons with quadriceps atrophy following surgery as well as those deconditioned athletes. Physicians should consider this information when prescribing a custom functional knee orthosis to an ACL deficient or ACL reconstructed knee person who plans to engage in strenuous activities.

Stephen H. Liu, MD

Volumetric Determinations With CAD/CAM in Prosthetics and Orthotics: Errors of Measurement.

Magnus Lilja, CPO, LMS and Tommy Öberg, MD, PhD *(p. 141)*

Purpose of the Work. During the next decade, CAD/CAM technique will probably become routine in prosthetics and orthotics, not only as a complement to manual techniques, but also in introducing new possibilities. However, even complex and sophisticated techniques have errors of measurement that must be considered. Such errors are of two principal kinds: systematic errors and random errors. **Subjects.** None. **Procedures.** In the present study, the Swedish CAPOD system has been evaluated with respect to volumetric determinations. Two types of reference objects were used for volume determinations: cylinders and amputation stump models. Three different sizes were examined for each type of object. Volume measurements with CAPOD were compared with volumes obtained by water immersion or mathematical calculation (cylinders only). **Results.** A constant, linear systematic error of +2.5% was found. Such an error can easily be corrected for. The random error, represented by the coefficient of variation, was 0.5%, which means that there is a theoretical possibility of detecting volume changes exceeding 1%. We considered the precision sufficient for clinical practice in prosthetics and orthotics. **Relevance to Veteran Population.** As the CAD/CAM technology is making entry into the fields of prosthetics and orthotics, it is of great importance that such systems are evaluated for accuracy and precision. The present study fulfills such a need.

Magnus Lilja, CPO, Lic Med Sci

An Animal Model and Computer-Controlled Surface Pressure Delivery System for the Production of Pressure Ulcers.

Richard Salcido, MD; Steven B. Fisher, PA-C; James C. Donofrio, PhD; Mark Bieschke, BS; Charles Knapp, PhD; Rongzhao Liang, MD, PhD; Edmund K. LeGrand, DVM, PhD; John M. Carney, PhD *(p. 149)*

Purpose of the Work. In order to better serve patients who develop bed sores, it is important that the medical

community learn more about their causes. Bed sores have been a persistent problem resulting in additional suffering and sometimes death in the patients who develop them. The economic costs to society are also high. Unfortunately, the scientific study of bed sores in a medical setting is often impossible due to their complex nature, patient considerations, and the difficulty in finding adequate control subjects. The development of an animal model for the study of bed sores avoids many of these problems and allows us to safely study how bed sores form. **Subjects/Procedures.** An animal model was developed for the study of bed sores. Rats were selected as the animal of choice because so much is known about their body chemistry and the effects of drugs on them. Computer control is used to achieve a highly controlled and closely monitored environment. **Results.** The computer-controlled model is described. Bed sores were consistently produced after five daily, 6-hour pressure sessions. The model is reliable and inexpensive and produces early bed sores over the hips, which allows us to study how bed sores form. The importance of the model and a review of the literature is discussed. **Relevance to Veteran Population.** Bed sores are an important health care problem with a major impact on large segments of the veteran population, including the aged and individuals with spinal cord injury.

James C. Donofrio, PhD

Experiments in Dysarthric Speech Recognition Using Artificial Neural Networks.

Gowtham Jayaram, MS and
Kadry Abdelhamied, PhD *(p. 162)*

Purpose of the Work. In order to facilitate the use of speech recognition technology by individuals with dysarthria, the recognition system must account for the high variability in dysarthric speech. The present study employs artificial neural networks for this purpose. **Subjects/Procedures.** Speech utterances produced by an individual with cerebral palsy were recorded and used to train and test two networks. The recognition rates of the networks were evaluated against the intelligibility ratings obtained by human listeners and also against the recognition rates obtained by the Introvoice commercial speech recognition system. **Results.** The networks successfully recognized dysarthric speech despite its large variability. The networks outperformed both the human listeners and the Introvoice commercial system. **Relevance to Veteran Population.** This study demonstrates the usefulness of neural networks in developing an effective speech recognition system for individuals with dysarthria. For dysarthric individuals with physical disabilities resulting from cerebral palsy, traumatic brain injury, stroke, and other neurogenic disorders, such a system is needed for computer access, environmental controls, and communication.

Kadry Abdelhamied, PhD

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Energy expenditure during ambulation in dysvascular and traumatic below-knee amputees: A comparison of five prosthetic feet

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Abstract—Recent advancement in prosthetic technology has led to the development of dynamic elastic response feet (DER), which are reported to store and release energy to facilitate gait. To date, there has been no objective evidence to suggest energy conservation while using these foot designs. The purpose of this study was to compare the energy expenditure of five commercially available prosthetic feet (SACH and four DER feet) in both the traumatic and dysvascular populations during level walking. Seventeen male subjects with below-knee amputation (nine traumatic and seven dysvascular) were tested for energy expenditure (Douglas Bag technique) during a 20-min walk while wearing each of the prosthetic feet. The DER prosthetic foot designs were not shown to reduce the energy cost (ml O_2/kg-m) or rate of energy expenditure (ml O_2/kg-min) compared to the SACH foot. Overall, the traumatic amputees had a similar oxygen consumption per meter traveled compared to the dysvascular amputees; however, the rate of energy consumption was much higher in the traumatic group. This increased rate was a function of the greater walking velocity employed by the traumatic subjects, made possible by their better physical fitness.

Key words: *amputees, energy cost, prosthetics.*

INTRODUCTION

Advancement in prosthetic technology during the last decade has led to the development of new prosthetic foot designs which have been termed Dynamic Elastic Response or DER feet. Compared to the traditional solid configuration of the SACH foot, the DER feet are reported to store and release energy in a manner that facilitates ambulation (1,2). The internal keel of these feet is designed to elastically deform under load bearing (thus storing energy) and then release when the amputee advances over the foot (3). This supposedly reproduces the energy absorption and generation characteristics of the normal foot and ankle in addition to providing better mobility. Theoretically, this would reduce the high energy cost of ambulation associated with this population.

Despite the reported energy storing and releasing properties of the DER feet, previous studies have failed to show statistically significant reductions in energy expenditure (i.e., oxygen consumption) when compared to the conventional SACH foot (4–7). To date, there has been only one report as to how the energy expenditure of the two primary populations of persons with below-knee (BK) amputation, due to trauma or vascular dysfunction, compares while walking with different prosthetic feet (7). In addition, biomechanical analysis during level walking has demonstrated that the SACH and DER feet have knee and hip power curves of comparable magnitude, implying that the metabolic energy consumption should be similar (8).

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These studies have been limited in sample size and have been inconsistent regarding the amputee population under investigation.

Compared to traumatic amputees, the dysvascular population tends to be less physically active, with a greater incidence of health-related problems (9). Thus, the traumatic amputee is more likely able to effectively compensate for the biomechanical limitations imposed by using a prosthesis than is the dysvascular amputee.

Studies have demonstrated that the energy expenditure per minute (O_2 rate) when walking at a self-selected comfortable speed, is equal between the traumatic amputees, dysvascular amputees, and subjects without amputation (10,11). Differences in walking velocity between these groups was responsible for the consistent O_2 rate. The energy cost (ml O_2/kg-m) for dysvascular amputees, however, has been reported to be greater when compared to traumatic amputees, with both groups having a greater energy cost than non-amputees because of a decrease in ambulation efficiency (11).

Little evidence exists that the energy "stored" in the DER feet is actually utilized to spare metabolic energy expenditure for either the dysvascular or traumatic amputee. The purpose of this study was to compare the effects of five different commercially available prosthetic feet (SACH and four DER feet) on the energy expenditure of walking in both the traumatic and dysvascular populations. This information could assist in providing a basis for prosthetic foot prescription for a patient with BK amputation (BKA), and would be beneficial in determining whether any of the DER feet provide the capacity for energy conservation.

METHODS

Subjects

Seventeen males with BKA, 10 traumatic and 7 dysvascular, participated in this study. The etiology of the dysvascular group was related to vascular disease secondary to diabetes. Subjects were recruited from the Long Beach Department of Veterans Affairs Medical Center (LBVA) STAMP program and from Rancho Los Amigos Medical Center Prosthetic Service.

Criteria for participation included independent community ambulation without use of an assistive device, and a history of compliance. All subjects consented to participate following explanation of the testing procedures and review of the informed consent form (approved by the LBVA Subcommittee for Human Studies). Following completion of the study, the subjects were able to choose one of the five feet tested to retain on a permanent basis.

Prosthetic Management

Five prosthetic feet were tested in random order for each subject: SACH,[1] Carbon Copy II,[2] Seattle Lite,[3] Quantum,[4] and Flex-Foot.[5] To ensure the fitting of appropriate foot components and keel, each manufacturer was provided with each subject's age, weight, height, contralateral shoe size, activity level, amputation level, and length of residual limb. The selection of the SACH foot heel wedge was based on the weight of the subject according to the guidelines of the developer. The appropriateness of each foot component/keel selection was then confirmed or modified at the time of prosthetic fitting.

The fit of each prosthesis was clinically optimized and reviewed by a team of three certified prosthetists. Alignment of the first foot followed established prosthetic principles. The Vertical Fabrication Jig[6] was used for subsequent alignments when more than the interchange of a foot-bolt was required.

Instrumentation

Each subject was fitted with three precordial electrocardiograph electrodes to monitor heart rate, a compression closing switch taped to the bottom of one shoe to record stride frequency (converted to cadence), and a harness with a telemetry system (**Figure 1**). A nose clip was placed on the subject to prevent nasal breathing. The harness was equipped with a mouthpiece attached to a T-segment with two one-way valves which allowed inspiration of ambient air and expiration into the collection bags (modified Douglas bags). A multiported system allowed attachment of multiple collection bags. A thermistor placed in the T-segment detected difference in temperature between the ambient air and expired air, allowing recording of respiration rate.

A level, 60.5 m outdoor track was used for the walking trials. Each meter of the track was marked for monitoring the distance traveled.

Respiration rate, heart rate, and stride frequency were recorded via telemetry on a strip chart recorder.[7] Gas analyzers were used to determine the carbon dioxide and oxygen content of the collected sample of expired air.[8] The temperature of the sampled air was monitored by a thermistor placed in the sample flow line. The volume of the collected expired air was measured by evacuating the col-

Figure 1.
Instrumentation used for energy cost testing. A multiported harness system collected expired air in modified Douglas bags during the 20-min walking session.

lection bag through a gas flow meter.[9] A mercury barometer was used to determine the atmospheric pressure at the time of testing.

Procedures

Subjects were given an accommodation period of 1 month to adjust to each prosthetic foot. Testing with the five prosthetic feet occurred at approximately 1-month intervals over a 5-month period. The testing procedures were identical for each session.

Body weight was recorded prior to each trial. Energy expenditure at rest was recorded after the subject had been seated for 30 min and fully instrumented for 5 min. Following the rest period, a self-selected free walk was completed. The duration of the walk was a minimum of 5 min and a maximum of 20 min. After the minimum 5 min, the subjects were allowed to terminate the testing when they felt unable to walk any longer.

Individual gas samples, heart rate, and respiration rate were recorded during the last 2 min of the 5 min rest period and at minutes 4 to 5, 9 to 10, 14 to 15, and 19 to 20 during walking. Stride frequency was also recorded during the walking trial. Distance traveled during each collection period, and the total distance walked were monitored by one of the investigators. Barometric pressure at the time of testing was recorded for use in calculating gas volumes.

Data Management

To calculate energy expenditure at rest and during walking, carbon dioxide content, oxygen content, temperature, and volume of the collected samples of expired air were used. Oxygen consumption values were converted to standard temperature, pressure, dry (STPD). Body weight (kg) was used to convert the oxygen consumption to ml of O_2 consumed per kg-min (rate of energy expenditure). For the walking data, body weight and velocity (m/min) were used to determine the ml of O_2 consumed per kg-m (energy cost per unit distance).

Cadence was calculated as twice the stride frequency. Stride length was calculated from the cadence and distance walked per minute.

The carbon dioxide produced and the oxygen consumed, as measured from the gas analyses, were used to calculate the respiratory exchange quotient (RQ) at rest, and the respiratory exchange ratio (RER) during walking.

Data Analysis

The data from the last collection period of each walking trial were used for comparisons between groups and prosthetic feet. Statistical analyses were performed using BMDP statistical software.[10] All data were analyzed for normality of distribution using the Shapiro and Wilk's W-statistic. Differences between prosthetic feet and groups for each parameter measured were determined either by a two-way analysis of variance (ANOVA) with repeated measures, or a Friedman's two-way ANOVA with repeated measures for those data not normally distributed. A 95 percent confidence level was used to determine statistical significance. A post-hoc Tukey test was used to find the significantly different comparisons.

RESULTS

Of the 17 subjects tested, 1 (a traumatic BKA) moved out of the area after completing four of the five sessions, and was therefore dropped from the analysis. Thus, the following data are from the remaining 16 subjects. The two groups were of similar age, height, and weight (**Table 1**).

As shown on **Table 2,** the heart rate (HR) during rest of the dysvascular group was significantly greater than that for the traumatic group (79 vs. 65 bpm; $p<0.05$). However, there was no significant difference in energy rate (oxygen consumption per min) at rest between groups or test days for all subjects. In addition, the RQ at rest did not vary between trials, days, or between the groups of amputees.

Eight of the nine traumatic amputees were able to complete the full 20-min test protocol. Only one of the dysvascular amputees was able to complete the 20-min trial. Failure to complete the 20-min walk was secondary to complaints of calf and/or anterior leg pain in the sound limb, or complaints of generalized fatigue. The traumatic group demonstrated a significantly greater average total walking time than the dysvascular group (18.8 vs. 10.3 min) as well as a significant increase (1.5 vs. 0.65 km) in distance traveled (**Table 3**).

During the walking trials, there was no statistical difference in velocity of ambulation between the five feet tested. However, the velocity of the traumatic group was significantly greater (82.3 vs. 61.7 m/min; $p<0.01$) than that of the dysvascular group when averaged across all foot conditions (**Figure 2**). Stride length did not vary with foot type, however, the traumatic group had a greater stride length (1.48 vs. 1.22 m; $p<0.01$) compared to the dysvascular group (**Figure 3**). Cadence was not significantly altered by the type of foot worn, nor by the etiology of the BKA. A trend toward decreased cadence in the dysvascular group was exhibited (100 vs. 110 steps/min) compared to the traumatic group (**Figure 4**).

During the last minute of the walking trials, the traumatic group had a lower heart rate than the dysvascular group; however, this was not statistically significant (113 vs. 116 beats/min), and respiratory rate (24 vs. 29 breaths/min) showed a similar response (**Table 4**). Heart rate and respiratory rate did not vary with the type of foot worn.

Among the prosthetic feet used, no significant changes in the rate of energy consumption were identified; however, the traumatic group had an energy rate (17.7 vs. 13.2 ml O_2/kg-min; $p<0.01$) that was statistically greater than the dysvascular group when averaged across all conditions (**Figure 5, Table 5**). The net energy cost (oxygen consumption per distance traveled) was not statistically different between the types of feet tested or between the groups of amputees (**Figure 6, Table 6**).

The walking data revealed a slightly higher, but not statistically significant, RER for the dysvascular group compared to the traumatic group (0.91 vs. 0.86), and the oxygen pulse (oxygen rate to heart rate ratio) was significantly greater (0.16 vs. 0.11 ml/kg-beat; $p<0.01$) in the traumatic group compared to the dysvascular group (**Table 4**). Foot type did not affect RER or the oxygen pulse ratio for either group.

Table 1.
Group characteristics: Mean (SD).

	Traumatic (N=9)		Dysvascular (N=7)		p-value
Age (yrs)	50.6	(15.6)	62.0	(8.3)	0.105
Height (cm)	175.1	(5.8)	177.2	(12.4)	0.662
Weight (kg)	81.2	(12.6)	85.2	(13.0)	0.537

Table 2.
Energy cost at rest: mean (SD).

	Traumatic		Dysvascular		p-Value
Heart rate (beats/min)	65.4	(10.8)	79.4	(15.4)	0.035*
Energy rate (ml O_2/kg-min)	3.5	(0.77)	3.1	(0.69)	0.232
Respiratory Quotient (RQ)	0.80	(0.10)	0.80	(0.11)	0.995

* Dysvascular group significantly greater than the traumatic group ($p<0.05$).

DISCUSSION

The altered mechanics of BKA gait has been well documented in the literature (4,8,12,13). To regain lost walking capability, the amputee exerts additional effort which is reflected in the increased oxygen consumption per meter traveled during gait (**Figure 6, Table 6**). Waters (11) reported this increased energy consumption to be 167 and 129 percent of normal for dysvascular and traumatic BKAs respectively. Comparable values were found in the current study. These increases reflect the demands on the remaining musculature. The lack of normal ankle mobility in loading response and single limb support necessitates compensatory gait patterns and muscle activity to provide stability and advancement over the foot (14). In the amputated limb, the large muscles controlling the hip and knee demonstrate more intense and prolonged elec-

Table 3.
Stride characteristics during energy cost testing, averaged for all feet tested*: mean (SD).

	Traumatic		Dysvascular		p-Value
Distance (km)	1.5	(0.43)	0.65	(0.41)	0.0009**
Time (min)	18.8	(3.3)	10.3	(5.8)	0.0015**
Velocity (m/min)	82.3	(16.9)	61.7	(8.7)	0.0065**
Stride length (m)	1.49	(0.20)	1.22	(0.14)	0.0068**
Cadence (steps/min)	110.3	(12.1)	100.8	(6.7)	0.0545

* No significant differences were found between feet for these variables.
** Traumatic group significantly greater than the dysvascular group ($p<0.50$).

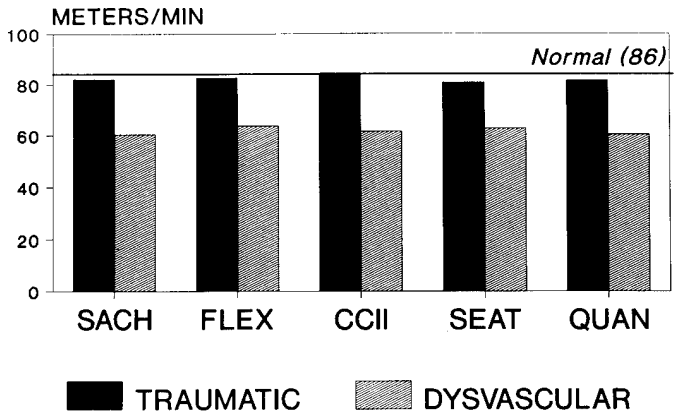

Figure 2.
Walking velocity (m/min) for both dysvascular and traumatic amputees for each foot tested. Flex=Flex-Foot, CCII=Carbon Copy II, Seat=Seattle Lite, Quan=Quantum. Normal data from (14).

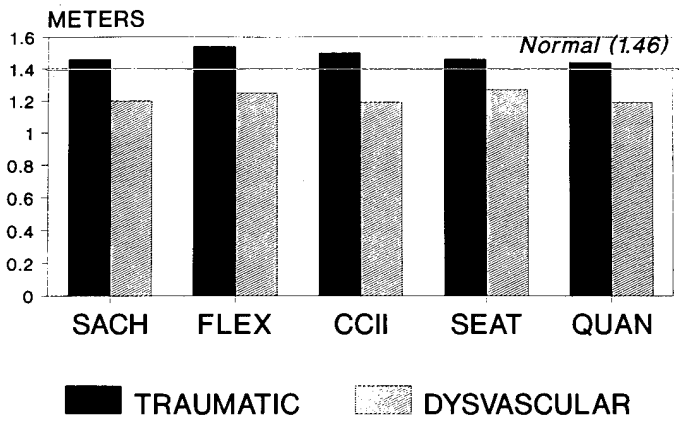

Figure 3.
Stride length (m) for both dysvascular and traumatic amputees for each foot tested. Flex=Flex-Foot, CCII=Carbon Copy II, Seat=Seattle Lite, Quan=Quantum. Normal data from (14).

tromyographic activity compared to those of persons without amputation and is consistent with the increased energy cost (4).

The recent designs of the DER prosthetic feet have focused on providing increased ankle mobility and reported energy return via a flexible internal keel to improve "push-off" mechanics. In addition to previous kinematic analyses that have found increased dorsiflexion mobility with the DER prosthetic feet compared to the SACH foot (4,15), indirect power calculations have implied energy conservation by these new prosthetic designs (8). The results of this study of direct energetics demonstrate that the energy cost and the rate of oxygen consumption were not influenced by prosthetic foot design, regardless of the etiology of amputation. This finding is in agreement with the work of Barth, Shumacher, and Sienko-Thomas (7) who also found no energy cost differences between six prosthetic feet in a much smaller population (N=6).

Given the small differences found in the current investigation for both energy cost and rate of energy expenditure between the five feet, combined with a relatively modest population (N=17), the statistical power of this analysis was quite low (0.13). Sample size determination found, however, that it would take approximately 70 subjects to show a statistical difference between prosthetic feet. This finding implies that the small difference detected in such an analysis would probably not be considered clinically significant.

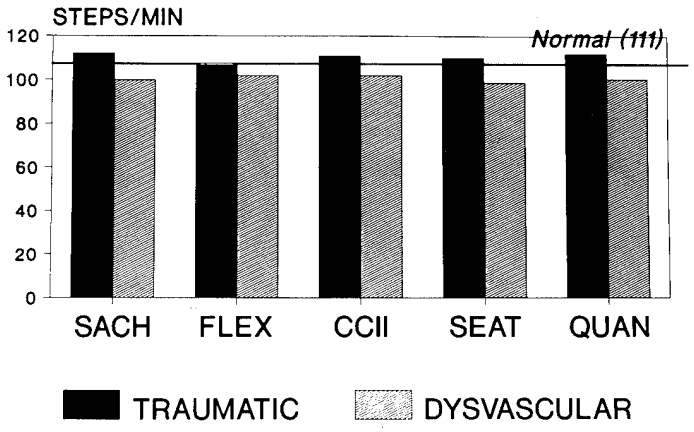

Figure 4.
Cadence (steps/min) for both dysvascular and traumatic amputees for each foot tested. Flex=Flex-Foot, CCII=Carbon Copy II, Seat=Seattle Lite, Quan=Quantum. Normal data from (14).

Previously reported data have indicated that BKAs have a slower gait velocity, decreased stride length, and decreased cadence compared to persons without amputations (4,6–8,11–13). Compared to the earlier data of Waters (11), both amputee groups in this study walked at greater unrestrained velocities (dysvascular: 64 vs. 45 m/min; traumatic: 82 vs. 71 m/min). Velocity for the traumatic group was close to the value of 86 m/min established by Perry (14) for male subjects without amputation. Foot type had no influence on velocity in either group.

The dysvascular amputees' slower velocity related to their less than normal stride length, as cadence, was not significantly different between groups (**Figures 3** and **4**). Since stride length was not affected by foot design, the shorter stride length in the dysvascular group suggests decreased strength compared to the traumatic amputees.

While the values for energy cost per meter traveled were similar between groups (**Figure 6, Table 6**), the near normal stride characteristics of the traumatic group were achieved at a greater rate of energy expenditure compared to their dysvascular counterparts (**Figure 5, Table 5**). This high rate of energy expenditure combined with a normal velocity, resulted in an energy cost (0.217 vs. 0.214 ml O_2/kg-m) similar to the more slowly ambulating dysvascular amputees (**Figure 6, Table 6**).

In contrast, Barth, Shumacher, and Sienko-Thomas (7) reported a significant difference in energy cost between their traumatic and dysvascular groups. While the energy cost values for the traumatic group in this study were similar to that of our data, their energy cost data for their dysvascular amputees were considerably less (0.214 vs. 0.130 ml O_2/kg-m). This was a function of the decreased velocity which was more pronounced in their dysvascular group. On the average, the dysvascular amputees in our study had a greater walking velocity compared to these previous results using a treadmill (45.0 vs. 61.7 m/min). This would account for the variability in energy cost results, as walking velocity has been shown to increase the rate and magnitude of loading, thus requiring increased muscular demand during loading response (16).

The increase in the rate of energy expenditure by the traumatic BKAs represents their capacity to walk faster, and reflects a higher level of physiological fitness. Despite the similar net energy cost between the two groups, the traumatic BKAs were able to cover a larger distance at a greater velocity, while the dysvascular amputees required more time to cover a comparable distance.

After 20 min of ambulation, the traumatic BKA subjects had a 147 percent greater rate of energy expenditure than normal (**Figure 5**). Compared to the traumatic amputee data of Pagliarulo (13), there was an increased rate of oxygen consumption in our traumatic group (17.7 vs. 15.5 ml O_2/kg-min). This is reflective of the relatively faster walking velocities that were evident in the current investigation (82.3 vs. 71.0 m/min). As expected, the mean net energy cost values for the traumatic amputees between these two studies was similar (0.217 vs. 0.218 ml O_2/kg-m).

The dysvascular group maintained a nearly normal energy rate (112 percent of normal) by walking at a slower velocity (83 percent of normal). This decrease in velocity in combination with a normal energy rate, however, resulted in a relatively high energy expenditure per meter traveled (136 percent of normal). The rate of energy expenditure for the dysvascular group also was higher than the rate previously reported by Waters (11) for this population (13.7 vs. 11.7 ml O_2/kg-min). As with the traumatic group, the difference between these values was a function of the increased walking velocity exhibited in this investigation (61.7 vs. 45.0 m/min).

The poor endurance of the dysvascular group was demonstrated by their inability to complete 20 minutes of walking, even at an energy expenditure rate only 12 percent greater than normal. Respiratory exchange ratios of 0.91 indicate the dysvascular subjects were working at anaerobic levels and support the subjective complaints of fatigue (17). Pain reported in the sound limb suggests systemic vascular compromise in these subjects. The similar, but slightly lower RER (0.86) of the traumatic group in-

Table 4.
Energy cost parameters collected during last minute of walking averaged across all feet tested*: mean (SD).

	Traumatic		Dysvascular		p-Value
Heart rate (beats/min)	113.7	(19.1)	116.4	(17.8)	0.749
Respiratory rate (breaths/min)	23.8	(6.8)	29.1	(7.7)	0.137
Oxygen pulse (ml/kg-beat)	0.16	(0.03)	0.11	(0.02)	0.002**
Respiratory Exchange Ratio (RER)	0.86	(0.06)	0.91	(0.07)	0.086

*No significant differences were found between prosthetic foot type for these variables.
** Traumatic group significantly greater than the dysvascular group ($p<0.01$).

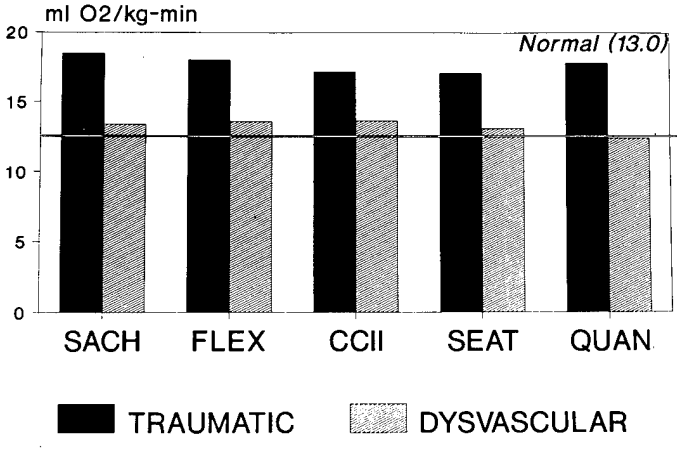

Figure 5.
Energy consumption rate (ml of oxygen per kg-min) for both dysvascular and traumatic amputees for each foot tested. Flex=Flex-Foot, CCII=Carbon Copy II, Seat=Seattle Lite, Quan=Quantum. Normal data from (11).

Table 5.
Energy consumption rate (ml O_2/kg-min): Mean (SD).

	Traumatic		Dysvascular	
SACH	18.48	(3.0)	13.41	(2.8)
Flex	18.04	(3.6)	13.61	(1.7)
Carbon Copy II	17.18	(3.6)	13.66	(2.7)
Seattle	17.08	(2.7)	13.10	(2.2)
Quantum	17.79	(3.5)	12.41	(2.3)
Average*	17.72**		13.23	

*Averaged across all feet tested.
**Traumatic > Dysvascular ($p<0.01$).

dicates this group of amputees was better able to tolerate the impairments imposed by their amputation than the dysvascular group. The higher oxygen pulse of the traumatic group (0.16 ml/kg-beat vs. 0.11 ml/kg-beat) further demonstrates their better physical conditioning compared to the dysvascular group (**Table 4**).

Endurance for walking at self-selected comfortable speeds was not improved with use of DER feet compared to the SACH foot. While the DER feet are reported to assist with energy return to the amputee to help reduce the physical demand of walking, neither group demonstrated changes in energy expenditure as a result of varying the type of prosthetic foot.

The DER feet were designed for the more vigorous amputees to facilitate running. Their emphasis was improved forward progression through greater ankle dorsiflexion. It was assumed that improved progression and push-off conserved energy. The findings of this study and others, imply that progression over the prosthesis and the initiation of swing limb advancement are not the prime causes of energy cost. Rather, peak energy cost mechanics occur elsewhere in the gait cycle.

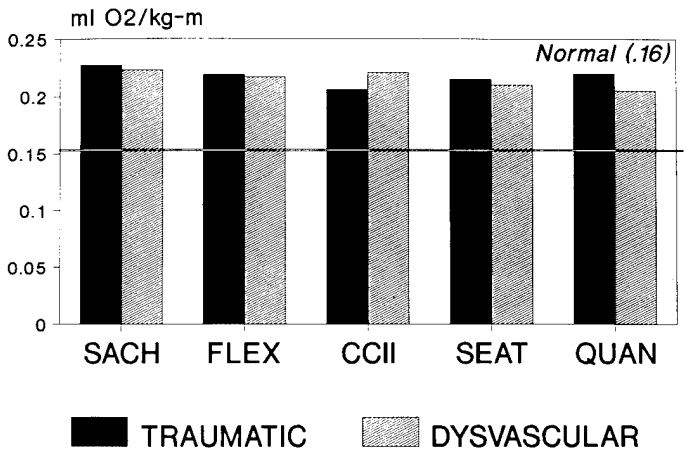

Figure 6.
Energy cost per distance traveled (ml of oxygen per kg-m) for both dysvascular and traumatic amputees for each foot tested. Flex=Flex-Foot, CCII=Carbon Copy II, Seat=Seattle Lite, Quan=Quantum. Normal data from (11).

Table 6.
Energy cost per distance traveled (ml O_2/kg-m): Mean (SD).

	Traumatic		Dysvascular	
SACH	0.227	(0.03)	0.223	(0.04)
Flex	0.219	(0.02)	0.215	(0.03)
Carbon Copy II	0.207	(0.03)	0.221	(0.02)
Seattle	0.215	(0.04)	0.211	(0.04)
Quantum	0.220	(0.03)	0.206	(0.03)
Average*	0.217		0.214	

*Averaged across all feet tested.

CONCLUSION

The DER prosthetic foot designs were not shown to reduce the physiologic demand of walking compared to the SACH foot. This was true for both the dysvascular group, for whom walking has been demonstrated to be a nearly anaerobic activity, and the traumatic amputees, who were working at a higher rate of energy expenditure. These data imply that the maximum muscular effort occurs at a time in the gait cycle which is not influenced by the DER foot mechanics. Future studies may incorporate a higher demand activity (other than free walking) to determine if energy conservation would be more evident.

ENDNOTES

1. Kingsley Manufacturing Co., Costa Mesa, CA.
2. The Ohio Willow Wood Co., Mount Sterling, OH.
3. Model and Instrument Development, Seattle, WA.
4. Hosmer-Dorrance Corp., Campbell, CA.
5. Flex-Foot Inc., Laguna Hills, CA.
6. Hosmer-Dorrance Corp., Campbell, CA.
7. Model 302, Astro-med, West Warwick, RI.
8. Oxygen Analyzer OM-11; Medical Gas Analyzer LB-2; Ventilation Measurement Module VMM series: Sensor Medics Corp., Anaheim, CA.
9. Collins motor #P-553-P, Warren E. Collins Inc., Braintree, MA.
10. BMDP Statistical Software Inc., Los Angeles, CA.

REFERENCES

1. Michael J. Energy storing feet: a clinical comparison. Clin Prosthet Orthot 1987:11(3):154–68.
2. Wing DC, Hittenberger DA. Energy-storing prosthetic feet. Arch Phys Med Rehabil 1989:70:330–4.
3. Edelstein JE. Prosthetic feet: state of the art. Phys Ther 1988:68:1874–81.
4. Torburn L, Perry J, Ayyappa E, Shanfield SL. Below knee amputee gait with dynamic elastic response prosthetic feet: a pilot study. J Rehabil Res Dev 1990:27(4):369–84.
5. Lehmann JF, Price R, Boswell-Bessette S, Dralle A, Questad K. Comprehensive analysis of dynamic elastic response feet: Seattle Ankle/Lite Foot versus SACH foot. Arch Phys Med Rehabil 1993:74:853–61.
6. Nielsen DH, Shurr DG, Golden JC, Meier K. Comparison of energy cost and gait efficiency during ambulation in below-knee amputees using different prosthetic feet: a preliminary report. J Prosthet Orthot 1988:I(I):24–31.
7. Barth DG, Shumacher L, Sienko-Thomas S. Gait analysis and energy cost of below-knee amputees wearing six different prosthetic feet. J Prosthet Orthot 1992:4(2):63–75.
8. Gitter A, Czerniecki JM, DeGroot DM. Biomechanical analysis of the influence of prosthetic feet on below-knee amputee walking. Am J Phys Med Rehabil 1991:70:142–8.
9. Stewart CPU, Jain AS. Cause of death of lower limb amputees. Prosthet Orthot Int 1992:16:129–32.
10. Waters RL, Perry J, Chambers R. Energy expenditure of amputee gait. In: Moore WS, Malone JM, eds. Lower extremity amputation. Philadelphia: W.B. Saunders Co., 1989:250–60.
11. Waters RL, Perry J, Antonelli D, Hislop H. Energy cost of walking of amputees: the influence of level of amputation. J Bone Joint Surg 1976:58A:42–6.
12. Winter DA, Sienko SE. Biomechanics of below-knee amputee gait. J Biomech 1988:21(5):361–7.
13. Pagliarulo MA, Waters R, Hislop HJ. Energy cost of walking of below knee amputees having no vascular disease. Phys Ther 1979:59:538–42.

14. Perry J. Gait analysis, normal and pathological function. Thorofare, NJ: Charles B. Slack, 1992.
15. Wagner J, Sienko S, Supan T, Barth D. Motion analysis of SACH vs. Flex-Foot in moderately active below-knee amputees. Clin Prosthet Orthot 1987:11(1):55–62.
16. Skinner SR. The relationship of gait velocity to the rate of lower extremity loading and unloading. Trans Ortho Res Soc 1980:5:273.
17. Astrand PO, Rodahl, K. Textbook of work physiology. 2nd ed. New York: McGraw-Hill Book Co., 1986:513.

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Gait initiation of persons with below-knee amputation: The characterization and comparison of force profiles

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Abstract—The purpose of this study was to characterize gait initiation of persons with leg amputation and determine whether prosthetic alignment was a critical parameter in the initiation process. Gait initiation was chosen for study because of the difficult neuromuscular demands placed on the body in negotiating the transition from stance to ambulation. In this investigation, ground reaction force data were collected on seven persons with below-knee amputation. These subjects underwent a series of gait initiation trials while varying prosthetic alignment. An analysis of the data demonstrated key elements in the gait initiation process, including the motion of the center of gravity in preparation for steady-state walking. Significant asymmetries in the force profiles of the residual and nonamputated limbs were also found; gait initiation forces were consistently higher for the prosthetic limb and the timings of maxima and minima were indicative of an intact limb preference. Relatively small changes in prosthesis alignment proved not to have statistically significant effects on generalized force parameters. This result is consistent with the findings of other studies that gait initiation is a motor program with certain invariant characteristics.

Key words: *amputation, biomechanics, gait, prosthesis, walking.*

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INTRODUCTION

Daily activities often require the transition from stance to ambulation, making the initiation of gait an important topic for investigation. Research has historically focused on the study of steady-state gait, leaving the complex process of initiating gait poorly understood. Studies of steady-state gait seem to be of limited usefulness in characterizing abnormalities, since many motor, sensory, and limb deficiencies can be hidden by the ability of individuals to produce similar level-walking kinematics with differing neuromuscular activity. Gait initiation, on the other hand, is a transient movement phenomenon and involves a more complicated integration of neural mechanisms, muscular activity, and biomechanical forces. The extensive involvement of the neuromuscular system implies that an understanding of gait initiation can potentially elucidate the etiology of different gait pathologies.

An early study of gait initiation conducted by Carlsoo (1) used electromyography (EMG) and force plate techniques to describe some of the major characteristics of the initiation process. This study first discovered that the initiation of walking is preceded by postural adjustments and muscle activity that direct the center of pressure (COP) over the swing limb before the transfer of weight to the stance limb. Subsequent work by other authors (2,3) described the neuromuscular aspects of gait initiation in attempts to understand the degree of central motor programming.

These studies suggested that central excitation of alphamotorneurons for both extensors and flexors during the gait initiation sequence remains largely unaffected by decreased spindle afferent output. This does not imply that fusimotor activity is not important, but that compensation may come from other somatic afferentation. A comprehensive study by Mann et al. (4) detailed the progression of events in gait initiation using force plate and EMG data. This study corroborated the findings of Carlsoo in demonstrating that the COP moves posteriorly and toward the swing limb in preparation for motion. Breniere et al. (5) introduced mathematical modeling of an inverted pendulum to explain the behavior of the COP, and to differentiate its behavior from that of the center of gravity (COG). It was this model that first demonstrated that the COP must move behind the COG in order for there to be forward motion. A recent article by Brunt et al. (6) used the invariance of the timing of events in the force and EMG profiles for gait initiation to postulate that gait initiation is a centrally programmed activity.

Persons with below-knee (BK) amputation can particularly benefit from research into gait initiation, for they must make the transition from stance to ambulation with diminished feedback from muscle spindle activity and ankle proprioceptive signals. Many studies have already investigated the steady-state walking characteristics of such persons (7–10). These studies used various combinations of force, EMG, and kinematic data to characterize the asymmetry found in their gait. Not surprisingly, persons with amputation exhibited lower average cadences, velocities, and stride lengths, while increasing the single limb support time on the intact limb. To date, the only work investigating gait initiation in persons with BK amputation, as well as in nondisabled persons is found in the studies by Nissan and Whittle (11) and Nissan (12). In these pilot studies, a protocol was established for comparing the gait initiation of subjects with amputation to that of nondisabled persons; it was found that the former loads the prosthetic limb during initiation less than the latter loads the analogous limb. It was emphasized that future work must be done to validate the use of generalized gait parameters as a clinical tool.

With their compromised muscular control and balance maintenance, persons with BK amputation must compensate for lost ankle function while ensuring safe, efficient gait initiation. Prosthetic alignment was hypothesized to be a critical parameter in this process. For this reason, this study was designed to demonstrate 1) the characteristic COP and ground reaction force (GRF) patterns of a person with BK amputation during gait initiation, and 2) the effect of prosthetic alignment on simple and interpretable force parameters. Understanding amputee gait initiation could potentially lead to amputee training techniques for improved negotiation of transient movement phenomena and to prosthesis modifications that could enhance stability during postural shifts.

METHODS

Subjects

Seven males with BK amputation were recruited for this study through the Special Team for Amputation and Mobility Preservation (**Table 1**). These subjects were selected based on reasonable ambulation skills and a willingness to participate in scientific research, were fully informed of the research protocol, and signed consent forms

Table 1.
Subject data.

Subject	Age	Weight (N)	Height (cm)	Amputation	Prosthesis Type	Socket Type
1	53	823	72	left	endoskeleton	PTB
2	68	1068	68	right	endoskeleton	PTS
3	82	912	72	left	endoskeleton	PTB
4	69	846	69	left	endoskeleton	Corset
5	50	765	69	right	endoskeleton	PTS
6	67	801	73	right	endoskeleton	PTB
7	62	1024	76	right	endoskeleton	PTB

approved by the University of California San Francisco Committee on Human Research. For the purposes of the investigation, each subject was provided an adjustable prosthesis. This involved the duplication of the current socket, the alignment of the prosthesis by an experienced prosthetist, and the acclimation of the subject to the new prosthesis. These prostheses housed modular adapters that permitted angular adjustment in the coronal and sagittal planes (**Figure 1**). Length adjustments could be achieved with the use of 2 cm spacers.

Measurement Protocol

Before beginning an experimental session, the subject was instructed in the general procedure and allowed to practice until comfortable with the protocol. Subjects were instructed to stand upright and balanced with one foot on each of two side-by-side force plates (Kistler Instrument Corporation, Amherst, NY). At the trigger of a light, the subjects initiated gait, and the collection of force data commenced, continuing for a period of 2.5 seconds. A trial consisted of six runs, three with the left limb leading and three with the right limb leading. In order to establish normative data, each subject underwent trials at normal initiation speed with the prosthesis aligned to the neutral position. Subsequent trials were conducted for each of six prosthetic adjustments. These included the positive and negative adjustment of the following alignment parameters: foot flexion (±5°), foot version (±5°), and leg length (±2 cm).

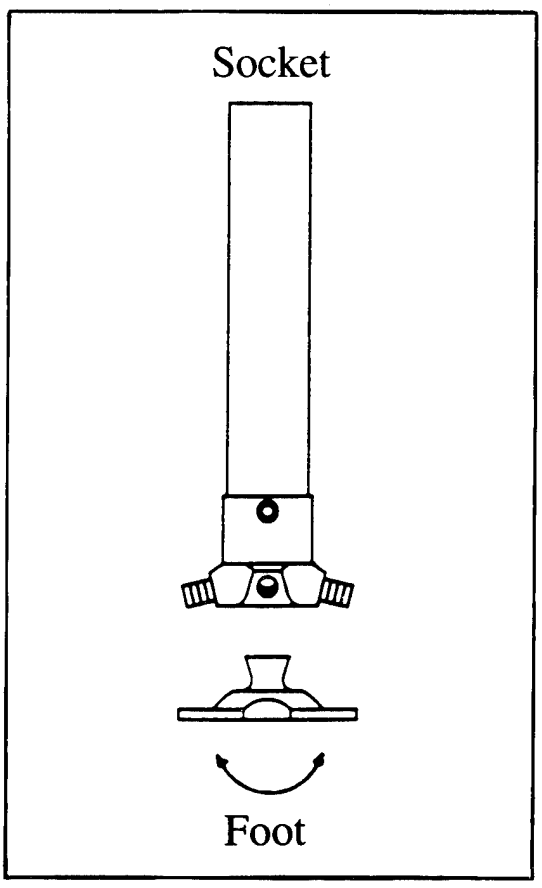

Figure 1.
The prosthetic limb pylon and the foot adaptor that allow for angular adjustment in the coronal and sagittal planes.

These alignment parameters were selected because of their importance from a clinical perspective and their ability to move points on the foot in all three directions of coordinate space. The magnitudes of the variations were large enough to cause imbalance and discomfort in the subjects, but small enough to be of clinical significance for the prosthetist performing the final minor adjustments to a prosthesis.

Data Collection and Analysis

The force data were collected at 200 Hz on a 486 IBM clone computer equipped with an analog-to-digital board and data acquisition software. The force plate transducer outputs were converted to force and COP data for each run. A FORTRAN program normalized the time to the period of initiation and the force to the body weight of the subject. The period of initiation was defined as the time between the onset of increased swing limb vertical forces and the termination of stance limb vertical forces, 1.25±0.15 (mean ± standard deviation) seconds in duration for the subjects studied. The program also identified the timings and magnitudes of the maxima and minima in the vertical, fore-aft, and medial-lateral force profiles for both the swing and stance limbs. The force magnitudes and their relative timings were tabulated for the 48 runs of each subject. The tabulated results became the data for a two factor ANOVA with repeated measures in order to identify differences produced by choice of leading limb (factor 1) and by alignment changes (factor 2). Significant effects were identified by p values less than 0.01.

RESULTS

The COP and the GRF profiles had similar shapes for all the subjects, regardless of the choice of leading limb. The excursion of the COP (**Figure 2**) began roughly centered between the two feet, usually favoring the side of the intact limb. It consistently moved posteriorly and toward

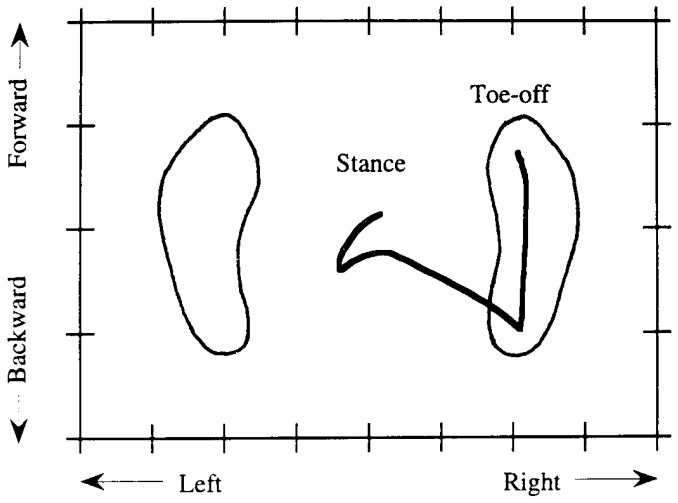

Figure 2.
The excursion of the COP during a typical gait initiation of a person with amputation. This particular example comes from subject 1 and demonstrates a left prosthetic limb lead.

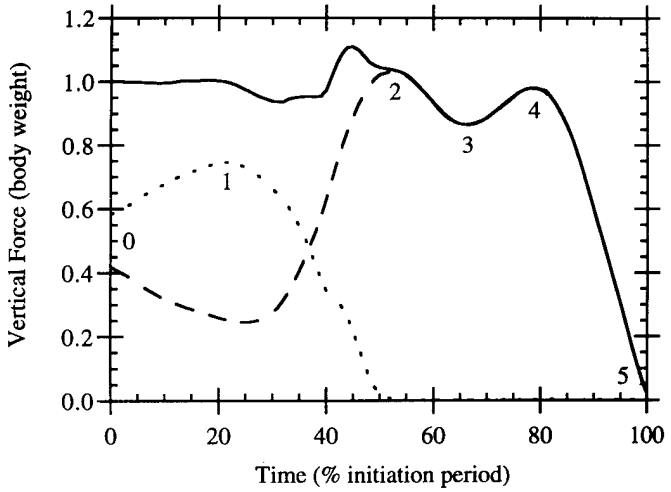

Figure 3.
The summation (solid line) of the contributions to the vertical force by the prosthetic swing (dotted line) and intact stance (dashed line) limbs during initiation for subject 2. The numbers 0 through 5 represent the significant maxima and minima in the gait initiation period, with 1 representing the simultaneous swing maximum and stance minimum.

the swing foot in a preparatory motion. It then reversed direction, moving laterally toward the stance foot heel. The COP finally traveled forward through the stance foot until toe-off. The vertical GRF profiles (**Figure 3**) consistently exhibited a shift in body weight to the swing limb prior to swing foot toe-off, the maximum shift occurring simultaneously with the COP direction reversal described above. The stance limb then began bearing the weight as it went into a single limb stance phase, which was characterized by a double-peaked force pattern also typical of steady-state gait. The summation of the vertical forces from the two limbs exhibited a profile that approximated body weight until single limb support and then followed the double-peaked pattern of single limb support. The fore-aft force profiles (**Figure 4**) demonstrated a net backward acting force exerted primarily by the swing limb prior to swing toe-off and by the stance limb during single limb support. The medial-lateral force profiles (**Figure 5**) demonstrated lateral acting forces, dominated initially by the swing limb and then by the stance limb.

The minima and maxima found on the swing and stance limb vertical GRFs (labeled in **Figure 3**) can be used to demonstrate quantitative differences between gait initiated by a prosthetic limb and gait initiated by an intact limb. Statistical analysis (**Table 2**) shows that the subject with amputation consistently loaded the intact limb more than the prosthetic limb, regardless of its role during initiation, whether as stance or swing limb. This tendency to

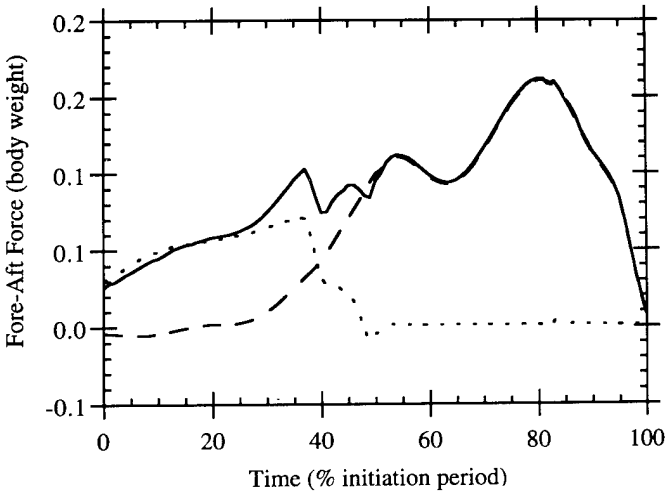

Figure 4.
The summation (solid line) of the contributions to the fore-aft force by the prosthetic swing (dotted line) and intact stance (dashed line) limbs during initiation for subject 2. Positive forces are in the backward direction.

load the intact limb more was true even during single limb support, a time when all the body weight was being supported by one limb. The timings of the force maxima and minima showed that the onset of prosthetic single limb support was delayed and that the second prosthetic stance

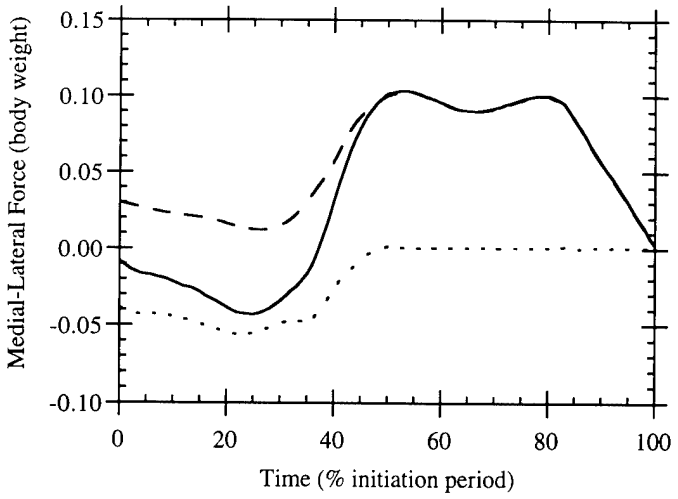

Figure 5.
The summation (solid line) of the contributions to the medial-lateral force by the prosthetic swing (dotted line) and intact stance (dashed line) limbs during initiation for subject 2. Positive forces are toward the left.

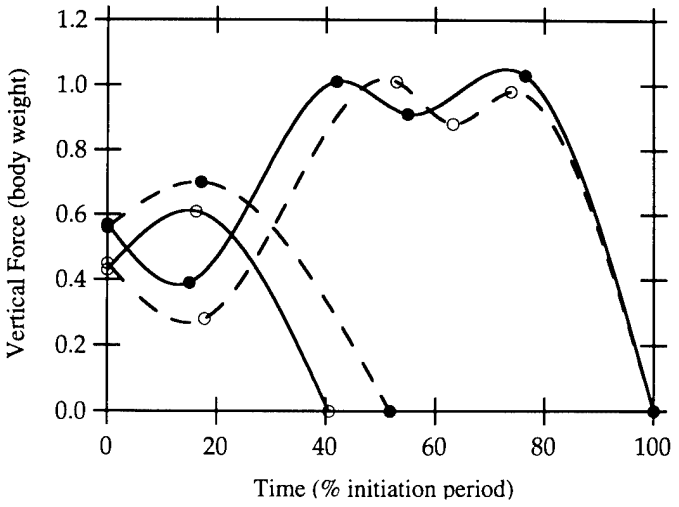

Figure 6.
A comparison between the vertical force profiles for the prosthetic (hollow dots) and intact (solid dots) limbs for the case when the prosthetic limb leads (solid line) and when the intact limb leads (dashed line). The data points are a statistical average of the maxima and minima taken from all subjects and all runs.

Table 2.
Comparison of vertical force maxima and minima (values in force normalized to body weight).

Event	Intact Limb	is	Prosthetic Limb	P Value
Stance 0	0.57 ± 0.09	>	0.45 ± 0.05	<0.0001
Stance 1	0.39 ± 0.16	>	0.28 ± 0.05	<0.0001
Stance 2	1.01 ± 0.03	>	1.01 ± 0.05	<0.1
Stance 3	0.91 ± 0.04	>	0.88 ± 0.05	<0.001
Stance 4	1.03 ± 0.07	>	0.98 ± 0.02	<0.0001
Swing 0	0.56 ± 0.05	>	0.43 ± 0.09	<0.0001
Swing 1	0.70 ± 0.05	>	0.61 ± 0.17	<0.001

Comparison of the timing of maxima and minima (values in percentage of gait initiation period).

Event	Intact Limb	is	Prosthetic Limb	P Value
Stance 1	15.0 ± 3.4	<	17.8 ± 3.3	<0.0001
Stance 2	41.8 ± 4.8	<	52.8 ± 5.9	<0.0001
Stance 3	54.8 ± 4.2	<	63.1 ± 4.8	<0.0001
Stance 4	76.4 ± 4.1	>	73.8 ± 4.2	<0.01
Swing 1	17.1 ± 3.2	>	16.2 ± 3.7	<0.01
Swing 2	51.7 ± 5.7	>	40.5 ± 5.0	<0.0001

peak occurred earlier, implying shorter prosthetic single limb support (**Figure 6**). Despite the inclusion of seven different patients and six prosthetic variations these statistics exhibited small variances, giving p values often much less than 0.0001.

Similar analysis was done for the fore-aft and medial-lateral GRFs (**Table 3**); only the absolute maximum was taken for each curve as the shapes were not as consistent between subjects. It was found that the swing limb peaks occurred earlier in the prosthetic than in the intact limb which is consistent with the vertical force findings. The swing limb fore-aft maximum was higher for the intact limb, and the stance limb medial-lateral force was higher for the prosthetic limb. Statistically significant findings were not as prevalent in the horizontal force profiles because of a higher degree of variability between subjects.

The variation of prosthetic alignment did not demonstrate significant effects on the generalized force parameters. The p values associated with these variables were often around 0.9, giving strong indication that no relationship existed for the selected alignment adjustments. With the seven subjects studied, there was an 80 percent chance of detecting an 8 percent change in either the force magnitudes or timings, assuming a 4 percent standard deviation of the test parameters and 95 percent confidence level for rejecting the null hypothesis. The 4 percent stan-

Table 3.
Comparison of horizontal force maxima (values in force normalized to body weight).

Event	Intact Limb	is	Prosthetic Limb	P Value
Stance - F/A	0.133 ± 0.045		0.127 ± 0.037	N.S.
Stance - M/L	0.078 ± 0.015	<	0.090 ± 0.021	<0.0001
Swing - F/A	0.065 ± 0.016	>	0.026 ± 0.024	<0.0001
Swing - M/L	0.075 ± 0.018		0.069 ± 0.023	N.S.

Comparison of the timing of maxima (values in percentage of gait initiation period)

Event	Intact Limb	is	Prosthetic Limb	P Value
Stance - F/A	78.0 ± 12.3		80.0 ± 6.0	N.S.
Stance - M/L	71.3 ± 7.7	>	66.9 ± 10.7	<0.02
Swing - F/A	25.3 ± 6.9	>	17.4 ± 6.8	<0.0001
Swing - M/L	23.1 ± 5.0	>	18.5 ± 4.3	<0.0001

dard deviation was determined using the within-subjects variation as an estimate of population standard deviation.

DISCUSSION

Center of Pressure Excursion

The fact that the COP moves posteriorly and toward the swing foot prior to swing toe-off has been well documented (1,4,5). Explanations for this phenomenon have been varied, but all lack completeness. The writings of Carlsoo and Mann et al. both postulated that COG follows COP over the swing foot in order to gather the momentum necessary to accelerate the body over the stance foot. They used the firing of the tibialis anterior and the suppression of the gastrocnemius as evidence that muscles are counteracting an initial fall backward, neglecting that these same muscles are involved in the acceleration of the body forward over the feet.

Mathematical analysis by Breniere et al. (5) used an inverted pendulum model to counter the explanations of previous authors by demonstrating that the COP must move backward in order for the COG to fall forward. This analysis was used to proclaim that a COP shift backward and toward the swing limb necessarily implies a fall forward and toward the stance limb.

Because the human body has multiple segments, however, we assert that the postural adjustment of body segments can result in motion in which COP and COG appear to move together. This would occur through a quasi-static process in which individual segments are accelerated and decelerated through small distances repeatedly over time. For example, shifting one's weight from one foot to the other through postural adjustments, provided it is done slowly, would not result in a dramatic oppositely moving COP shift and would in fact move both the COP and COG through the same distance.

Performing a simple force balance can help differentiate the behavior of COG from COP for gait initiation. While two plates registered forces independently in our experiment, the addition of these two forces indicated the net force acting on the body, which served to accelerate the COG. The summation of the fore-aft force profiles demonstrates that the net force acting on the body is always in the forward direction, dominated initially by the swing push-off and finally by stance push-off. As there is no part of the cycle which demonstrates negative fore-aft forces (posteriorly acting forces), there cannot be backward acceleration of the COG; even a quasi-static backward motion would have resulted in small negative initial forces followed by near zero fluctuating forces as the body moved in a controlled manner. Similarly, the summation of the medial-lateral force profiles showed swing limb-dominated lateral forces at the beginning of initiation, indicating that the COG was being pushed toward the stance limb from the start, to be redirected later in the cycle toward the swing limb. It is this oscillatory motion that keeps the COG controlled during bipedal forward motion.

The most interesting aspect of this COG motion is that it had always been previously explained by a falling mechanism, such as in the case of the inverted pendulum model. This concept is countered by the vertical force composite, which showed that prior to single limb support, the net vertical force approximates body weight fluctuating between values above and below body weight. In many of the subjects, this oscillation was even dominated by forces greater than body weight, indicating that there is no clear free-fall. If anything, there was a tendency to accelerate the body upward by the action of the stance foot on the floor. It is not until single limb support that one finds the characteristic dip of falling accelerations. It seems likely that preparatory gait initiation is a controlled forward/lateral push by the swing foot that puts COG and COP in the positions required to have a stance foot-assisted free fall.

Gait Initiation Asymmetry

A preliminary study by Nissan and Whittle (11) provided normative data on gait initiation force profiles for nondisabled subjects. A comparison of the intact limb vertical force maxima of our subjects to the maxima of their nondisabled subjects reveals that persons with amputation tend to have less pronounced peaks during single limb support (**Table 4**). In a subsequent study by Nissan (12), gait initiation of persons with amputation was compared to nondisabled gait initiation, and this difference in peak vertical force during single limb support was reported.

Our study, however, was the first to compare the profiles from both limbs on the same patient, gaining the statistical power of having paired data and enabling significant asymmetries to be found. It was found that persons with amputation consistently bear more weight on the intact limb whether it serves as the swing or stance limb, which has interesting implications on the postural adjustments necessary to keep the COG moving straight forward. Higher forces were found for the intact limb even during the single limb stance phase of initiation; this implies that the muscles of the intact limb facilitated higher upward accelerations, both in the deceleration of the falling COG and the acceleration of the upward moving COG. Possible reasons for not loading the prosthetic limb include a natural consequence of decreased proprioceptive feedback, a compensation for muscular deficiencies, and a result of pain and discomfort experienced during training.

Asymmetries were also found in the timing of events in the gait initiation cycle of persons with amputation. The loading of the prosthetic stance limb was significantly delayed when compared to the loading of the intact limb. It seemed as though gait initiation of persons with amputation was designed to minimize time spent on the prosthetic limb, which corroborates the findings in steady-state gait studies. The existence of gait asymmetries raises the question whether such a person should ideally be exhibiting symmetrical characteristics when his or her body is necessarily asymmetrical. Design can restore geometric symmetry, but because muscle function cannot be completely replaced by a prosthesis, compromises are necessary. One indication that symmetry might be a reasonable goal of design and training is that the work of Lewallen et al. (13) showed that children with amputation achieved good symmetry in joint kinematics despite physical asymmetry through slower velocities, smaller steps, and increased double limb support.

Prosthetic Alignment

After identifying the force characteristics of gait initiation in subjects with BK amputation, a goal of this research was to identify the effects of prosthetic alignment on these characteristics. Because gait initiation force profiles were remarkably reproducible, they could be almost completely characterized by their maxima and minima. A statistical analysis showed us that our prosthetic alignment variations produced no statistically significant effect on the generalized force parameters. This result is partially explained by the finding by Zahedi et al. (14) that persons with BK amputation can tolerate a great deal of variability in the "optimal" alignment. These results may also indirectly tie into the work of other authors (6,15) who found that gait initiation seems to be part of a central program and contains certain invariant characteristics, such as the timing of muscle and force events. Our experienced prosthesis wearers may have such patterned initiation processes that anything short of drastic alterations would have minimal effects. This does not imply that alignment is not important for the training of patients for efficient gait.

Table 4.
Comparison of the vertical force maxima found in this study and in the literature (values in force normalized to body weight).

Event	Nissan & Whittle (Normals)	Nissan (Intact / Prosthetic Limb)	Our Study (Intact / Prosthetic Limb)
Swing 1	0.76 ± 0.09	0.66 ± 0.08 / 0.60 ± 0.15	0.70 ± 0.05 / 0.61 ± 0.17
Stance 2	1.11 ± 0.05	0.98 ± 0.02 / 1.03 ± 0.04	1.01 ± 0.03 / 1.01 ± 0.05
Stance 3	0.89 ± 0.04	0.91 ± 0.04 / 0.90 ± 0.02	0.91 ± 0.04 / 0.88 ± 0.05
Stance 4	1.10 ± 0.06	0.98 ± 0.02 / 0.99 ± 0.03	1.03 ± 0.07 / 0.98 ± 0.02

CONCLUSIONS

The study of the forces involved in gait initiation has shown that walking is precipitated by a forward lateral push of the COG by the swing limb in preparation for an assisted fall by the stance foot. Persons with amputation perform this process asymmetrically, mostly in their tendency to weight the intact limb as much and as long as possible. The force characteristics of the gait initiation period seem not to be sensitive to small but significant alignment variations, especially with subjects whose gait patterns have long been established. These results do seem to indicate that the majority of effort should be placed at the early stages of gait training during which the asymmetries are established and walking patterns are set.

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REFERENCES

1. Carlsoo S. The initiation of walking. Acta Anat 1966:65(1):1–9.
2. Cook T, Cozzens B. Human solution for locomotion. III. The initiation of gait. In: Herman RM, Grillner S, Stein PSG, Stuart DG, eds. Neural control of locomotion. New York: Plenum Press, 1976:65–76.
3. Herman R, Cook T, Cozzens B, Freedman W. Control of postural reactions in man: the initiation of gait. In: Klopsteg PE, Wilson PD, eds. Control of posture and locomotion. New York: Plenum Press, 1973:363–88.
4. Mann RA, Hagy JL, White V, Liddell D. The initiation of gait. J Bone Joint Surg Am 1979:61(2):232–9.
5. Breniere Y, Do MC, Sanchez J. A biomechanical study of the gait initiation process. J Biophys Med Nucl 1981:5(4):197–205.
6. Brunt D, Lafferty M, Mckeon A, Goode B, Mulhausen C, Polk P. Invariant characteristics of gait. Am J Phys Med Rehabil 1991:70(4):206–12.
7. Breakey J. Gait of unilateral below-knee amputees. Orthot Prosthet 1976:30(3):17–24.
8. Robinson JL, Smidt GL, Arora JS. Accelerographic, temporal, and distance gait: factors in below-knee amputees. Phys Ther 1977:57(8):898–904.
9. Suzuki K. Force plate study on the artificial limb gait. J Jpn Orthop Assoc 1972:46:503–16.
10. Winter DA, Sienko SE. Biomechanics of below-knee amputee gait. J Biomech 1988:21(5):361–7.
11. Nissan M, Whittle MW. Initiation of gait in normal subjects: a preliminary study. J Biomed Eng 1990:12:165–71.
12. Nissan M. The initiation of gait in lower limb amputees: some related data. J Rehabil Res Dev 1991:28(2):1–12.
13. Lewallen R, Dyck G, Quanbury A, Ross K, Letts M. Gait kinematics in below-knee child amputees: a force plate analysis. J Pediatr Orthop 1986:6:291–8.
14. Zahedi MS, Spence WD, Solomonidis SE, Paul JP. Alignment of lower-limb prosthesis. J Rehabil Res Dev 1986:23(2):2–19.
15. Breniere Y, Do MC. When and how does steady state gait movement induced from upright posture begin? J Biomech 1986:19(12):1035–40.

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The prediction of metabolic energy expenditure during gait from mechanical energy of the limb: A preliminary study

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Abstract—Measurements of metabolic energy consumption and free-walking velocity were recorded for four persons with trans-femoral amputation with variations of prosthesis mass and mass distribution. Hot-film anemometers, rate gyroscopes, and a force platform were used to measure prosthetic limb segment velocities and ground reaction forces. Metabolic energy consumption for the nine configurations of mass and mass distribution averaged 1.177 cal/kg/m with a standard deviation of ±0.052 cal/kg/m. Two measures of mechanical work of the amputated extremity, one based on power developed across joints (W_1) and the other based on changes in energy of the body segments (W_2), were computed to be 0.162 ±0.014 and 0.175 ±0.025 cal/kg/m, respectively. A linear regression model led to rejections of both W_1 and W_2 as predictors of metabolic energy expenditure of the amputee at a significance level of 0.05.

Key words: *amputation, energy expenditure, gait analysis, rehabilitation.*

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INTRODUCTION

A major problem in the rehabilitation and reintegration of persons with amputation has been identified as the high metabolic energy cost of ambulation (1,2). This is particularly so for dysvascular trans-femoral (AK) amputees who approach anaerobic conditions during ambulation (3). If specified mechanical energy measurements reliably and accurately predicted metabolic energy expenditure, several benefits would be realized in regard to optimizing lower-limb prosthesis performance. Since prostheses can be designed in terms of mechanical variables, changes in design would be reflected in changes in metabolic energy expenditure. The establishment of a metabolic energy predictor, based on mechanical measurements, would also allow the energy aspects of gait to be studied through computer simulation. Instantaneous sources of energy expenditure during the gait cycle could be identified, and such analysis could be used in rehabilitation of pathological gait.

In a review of the measurement of mechanical energy associated with human movement, Winter notes that consensus is lacking in regard to the best method of calculating mechanical energy expenditure (4). Methods can be grouped into three categories. The first and second categories include work calculations based on power developed across joints and on changes in the energy of the body segments, respectively. In the third category, a separate energy

for the body center of mass and energy of the limbs relative to the center of mass are calculated (5). Although mechanical work is always less than the metabolic energy expenditure, since many physiological muscle processes are not represented in the model, mechanical work measures might be used to predict metabolic energy expenditure.

Human locomotion is accomplished by the exertion of muscle moments across the joints. Direct measurement of muscle force is impractical, but statically equivalent moments at the joints can be computed from ground reaction forces and limb segment accelerations. The muscle work of a rigid, linked model of N segments can be computed by:

$$W_1 = \int_{t_1}^{t_2} \sum_{i=1}^{N-1} M_{i,i+1}\left(\dot{\theta}_{i+1} - \dot{\theta}_i\right) \qquad [1]$$

where $M_{i,i+1}$ is the muscle moment acting between body segments i and i+1, and $\dot{\theta}_{i+1}$ and $\dot{\theta}_i$ are the angular velocities of segments i+1 and i, respectively. Only single joint muscles are modeled by Equation 1 (6). In addition, isometric contractions, elastic energy storage in muscles, and co-contractions of antagonistic muscles are not modeled, and are substantial limitations. Work calculated by Equation 1 is positive for concentric contraction and negative for eccentric contraction. The metabolic energy cost is always positive. A measure of metabolic energy expenditure associated with muscle work of Equation 1 is proposed:

$$W_1 = \int_{t_1}^{t_2} \sum_{i=1}^{N-1} \left| M_{i,i+1}\left(\dot{\theta}_{i+1} - \dot{\theta}_i\right) \right| dt \qquad [2]$$

Aleshinsky concluded W_1 is the most promising measure of mechanical work (7).

A measure of mechanical work based on changes in the energy of the leg can also be formulated. The total energy of the leg is the sum of the potential plus kinetic energies of the thigh and shank (including the foot). The instantaneous energy "rigid-body" model for motion in the sagittal plane is:

$$\begin{aligned} E_{leg} = {} & m_s g z_{cms} + \tfrac{1}{2} m_s\left(\dot{x}_{cms}^2 + \dot{z}_{cms}^2\right) + \tfrac{1}{2} I_s \dot{\theta}_s^2 \\ & + m_t g z_{cmt} + \tfrac{1}{2} m_t\left(\dot{x}_{cmt}^2 + \dot{z}_{cmt}^2\right) + \tfrac{1}{2} I_t \dot{\theta}_t^2 \end{aligned} \qquad [3]$$

where m_s, m_t are the segment masses, z_{cms}, z_{cmt} are the heights of the segment centers of mass above a defined reference, $\dot{x}_{cms}$, $\dot{x}_{cmt}$, $\dot{z}_{cms}$, $\dot{z}_{cmt}$ are the velocity components of the segment centers of mass, Is, It are the rotational inertias about the segment centers of mass and θ_s, θ_t, are the angular velocities of the segments. For positive metabolic work only, the work measure proposed can be represented by:

$$W_2 = \int_{t_1}^{t_2} \left| \frac{dE_{leg}}{dt} \right| dt \qquad [4]$$

The metabolic energy expenditure during gait for the entire organism will be greater than that for the trans-femoral prosthetic limb. However, since the energy expenditure of persons with trans-femoral amputation is significantly increased above normal (1), it was hypothesized that metabolic energy expenditure by the subject may be adequately predicted by one of these measures of mechanical energy of the limb.

METHODS

Four male subjects with traumatic trans-femoral amputation participated in this study (**Table 1**). All subjects were accustomed to wearing a prosthesis and did not suffer from residual limb pain, swelling, or pressure sores. Each subject signed a consent form approved by the University of California, San Francisco, Human Studies Committee. A lightweight prosthesis, consisting of a Teh Lin (Daw Industries, Inc., Eden Prairie, MN) graphite single-axis knee unit with adjustable constant friction and an adjustable internal extension assist, a Teh Lin graphite pylon and foot adapter, and a SACH foot, was constructed for each subject. Also, a lightweight polyester resin duplicate of each subject's current total-contact suction socket was fabricated. The addition of a swing phase control unit to the prosthesis was unnecessary because only free-walking velocity was studied. This also reduced the weight of the prosthesis. No mandate on footwear was imposed.

Instrumentation was developed to evaluate both W_1 and W_2. Calculation of W_2 required measurement of segment masses and the velocity of the centers of mass, as well as the inertias and angular velocities. In addition, measurement of the floor reaction forces was necessary to estimate the hip moment (M_h) for calculation of W_1 (**Figure 1**).

The mass, center of mass location, and inertia of each subject's residual limb were calculated by discretizing it into 2-cm thick elliptical slices. Measurements of the an-

Table 1.
Anthropometric data.

Subj.	Age	Body Mass Length (kg)	Height (m)	Residual Limb Length (m)	Floor-to-Knee Axis (m)	Combined Residual Limb and Socket Parameters M (kg)	Y_{cm} (m)	I_{cm} (kg-m²)	Combined Prosthetic Shank and Shoe Parameters M (kg)	X_{cm} (m)	Y_{cm} (m)	I_{cm} (kg-m²)	Combined Normal Shank and Foot Parameters M (kg)	Y_{cm} (m)	I_{cm} (kg-m²)
1	63	77.5	1.74	.260	.514	3.09	.220	.0540	1.65	.023	.325	.0611	3.86	.249	.103
2	58	81.8	1.70	.290	.551	4.71	.204	.0583	1.75	.035	.350	.0727	4.39	.239	.131
3	66	79.3	1.79	.270	.533	4.89	.265	.0855	1.69	.030	.327	.0709	3.97	.239	.121
4	68	74.5	1.79	.370	.503	4.45	.215	.0682	1.73	.029	.329	.0615	4.17	.238	.116

M = mass; Y_{cm} = center of mass coordinate in distal direction (reference is knee axis); X_{cm} = center of mass coordinate in interior direction (reference is knee axis); I_{cm} = Inertia about an axis perpendicular to the sagittal plane and located at the center of mass.

terior-posterior and medial-lateral dimensions of each slice were used to estimate major and minor axes of the ellipse. An average uniform density of 1050 kg/m³ was assumed (8). The center of mass of the prosthetic leg was determined by suspending it from two points and locating the intersection of the vertical lines through the support points. The inertia of the prosthetic leg was predicted from the measured period of oscillation when it was suspended at the knee axis. The center of mass and inertia of the thigh portion of the prothesis were determined in the same manner.

Translational velocities of the limb segments were measured using hot film anemometry. Dual sensor, x-configuration (TSI 1240-20, TSI Inc., St. Paul, MN) probes enabled velocity determination in the sagittal plane. Probes operated at 350° F, and were calibrated to yield an air velocity vector in polar coordinates that was determined uniquely in a 90° range. Sun, et al. assessed the accuracy of the hot film measurements to be ±0.1 m/s and ±10° at speeds greater than 0.25 m/s (9). Pendulum calibration methods have since been refined to yield ±5° accuracy in angle measurement for speeds greater than 0.25 m/s.

Miniature rate gyroscopes (United States Time Corp. #400) were used to measure the angular velocities of the thigh and shank. These units have a nominal full scale range of 400°/s, are linear within ±1.0 percent of full scale, and weigh 120 g. **Figure 2** shows the instrumented prosthesis. Identical hot film probes and rate gyroscopes were attached to the thigh and shank.

Ground reaction forces were recorded with a Kistler piezoelectric, multi-component, measuring platform (type 9261A, Kistler Instrument Corp., Amherst, NY). Precision and accuracy given by the manufacturer, are ±2 percent of full scale for force components.

Walking trials were conducted in a 12-m long gait laboratory. The configurations of distribution of mass on the prosthesis that were tested were 0, 1.70 kg, 2.84 kg, and 3.97 kg of added mass located 17, 25, and 33 cm distal to the knee axis. Configurations were tested in a random order, and two trials were recorded for each configuration. The prosthetic knee friction setting was adjusted between trials, and data were recorded after the subjects felt accustomed to a new configuration. A starting location that allowed at least four gait cycles before stepping on the force plate was selected. Data were recorded beginning with the first step.

All data were recorded on an IBM PC/AT using the ASYSTANT (Asyst Software Technology, Inc., Rochester, NY) data acquisition software package. Subjects wore a compact backpack (1.0 kg) which housed the hot film and rate gyroscope electronic units, and a 15 m cable linked the backpack to the computer. Signals were sampled at 150 Hz and collected for 10 seconds.

The work measures were calculated for the gait cycle starting when the prosthetic heel contacted the force plate. The anemometers measured the direction of the velocity vector relative to a reference frame fixed to the leg. The linear velocities in earth-fixed coordinates were deter-

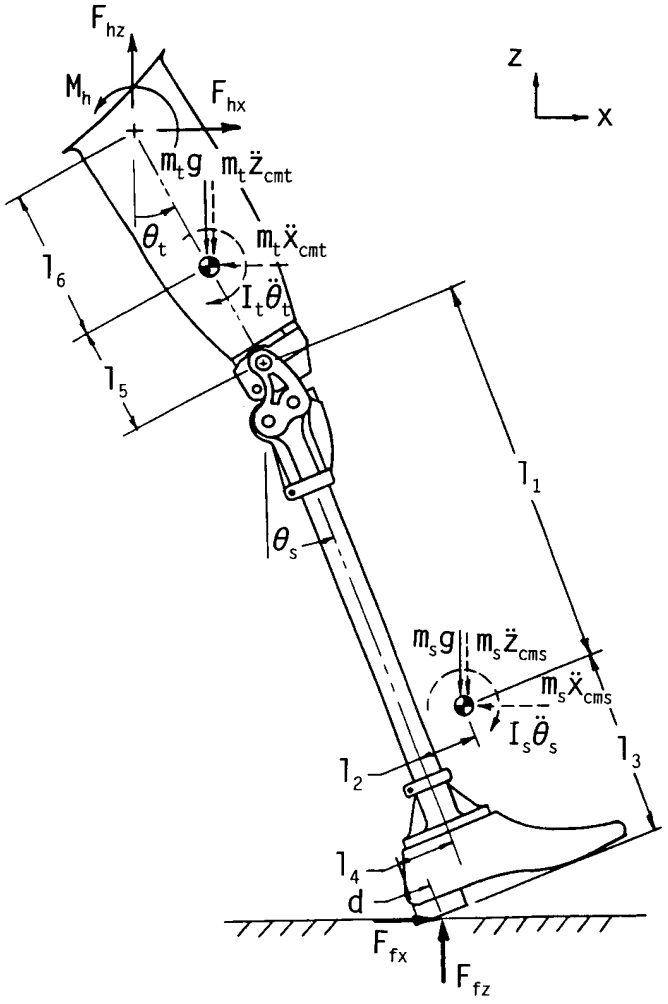

Figure 1.
Free-body diagram showing the forces and moments acting on the prosthetic leg. The moment at the hip is calculated from the equation:

$$M_h = I_t\ddot{\theta}_t + I_s\ddot{\theta}_s + m_s\ddot{X}_{cms}$$
$$(l_1\cos\theta_s - l_2\sin\theta_s + l_6\cos\theta_t + l_5\cos\theta_t)$$
$$+ m_s(\ddot{Z}_{cms} + g)(l_1\sin\theta_s + l_2\cos\theta_s + l_6\sin\theta_t + l_5\sin\theta_t)$$
$$+ m_t\ddot{X}_{cmt}l_6\cos\dot{\theta}_t + m_t(\ddot{Z} + g)\,l_6\sin\theta_t$$
$$- F_{fx}(l_1\cos\theta_s - l_2\sin\theta_s + (l_4 + l_2 - d)$$
$$\sin\theta_s + l_3\cos\theta_s + l_6\cos\theta_t + l_5\cos\theta_t)$$
$$-F_{fz}(l_1\sin\theta_s + l_2\cos\theta_s + (-l_4 - l_2 + d)$$
$$\cos\theta_s + l_3\sin\theta_s + l_6\sin\theta_t + l_5\sin\theta_t)$$

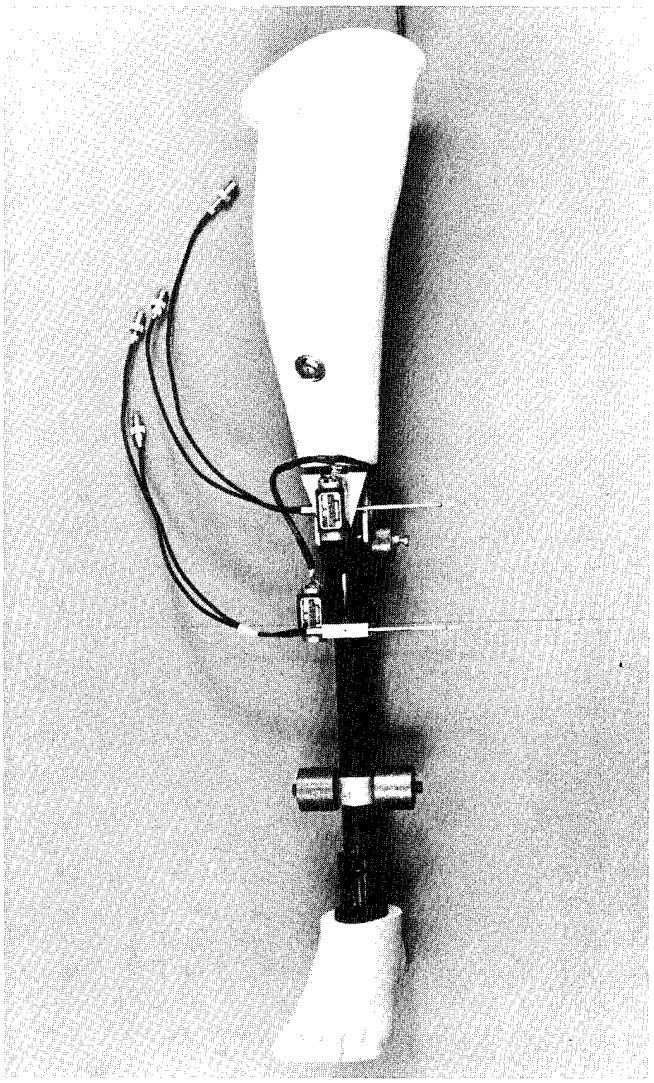

Figure 2.
Prosthesis instrumented for measurements of mechanical work. Two horizontal hot film probes used for linear velocity measurement are located at "knee" level and several inches below. The small rectangular box attached to each probe is a rate gyroscope used for angular velocity measurement.

mined by transformation using the orientation of the limb segments calculated by integrating the angular velocities. The velocity of the center of mass of a segment was calculated from the linear velocities at the anemometer and the segment angular velocity. The height of the center of mass was computed by integrating the vertical velocity. Energy was computed from the measured and computed data. Linear and angular accelerations of the thigh and shank were required to compute the hip moment (**Figure**

1 and Equation 2). Accelerations were calculated from the derivative of cubic splines fit to the linear and angular velocity data in a least squares sense (10). W_1 and W_2 were normalized by body weight and distance traveled to facilitate comparison with metabolic energy values.

The precision of W_1 and W_2 was studied by superimposing normally distributed, zero mean, stochastic errors on measured velocities and forces from a sample gait trial and recalculating work parameters 100 times.

Metabolic energy expenditure measurements were conducted in a 50-meter long straight and level corridor using indirect calorimetry by collecting exhaled gases in a 60-liter Douglas bag. Only free-walking velocity trials were investigated. The subject was given sufficient time to become accustomed to the prosthesis changes before the experiments. Total test time was approximately 5 minutes consisting of a 3-minute warm-up period to ensure that the muscles did not utilize anaerobic sources of energy (11), 1 minute to prime the airways, and 1 minute of exhaled gas collection. The order of test configurations was random, and three consecutive replications of the test were recorded at each configuration. Fatigue was minimized by allowing subjects to rest during the changing of weights.

The oxygen content of the exhaled gas was measured with a blood gas analyzer (Instrumentation Laboratory, IL 13040). A spirometer (RamAir 9200) was used to measure gas volume. Energy values were calculated from the formula recommended by Durnin and Passmore (11) and converted to units of cal/kg-m.

RESULTS

A summary of the data is given in **Table 2.** The average metabolic energy expenditure for the nine configurations of mass and mass location was 1.177 ±0.052 cal/kg/m. This compares with averages of 0.162 ±0.014 cal/kg/m and 0.175 ±0.025 cal/kg/m for W_1 and W_2, respectively. Units of cal/kg/m can be converted to mlO_2/kg/m by dividing by 5.

A linear regression relating metabolic energy expenditure to added mass, mass location, and W_1 or W_2, with tests for differences in subjects, was examined (12). All variables were rejected as predictors of metabolic energy expenditure at a significance level of 0.05. W_1 was calculated with a precision of ±0.0028 cal/kg-m (95 percent confidence) based on computations with random errors superimposed on measured variables from a sample gait trial. The precision of W_2 was ±0.0054 cal/kg-m (95 percent confidence). Linear regression of W_2 on W_1 yielded no relationship (p=0.18).

DISCUSSION

Previous investigators have examined the relation of metabolic energy consumption to mechanical work measures (13,14). Ralston and Lukin reported that W_2 correlated with metabolic energy consumption for changes in the total energy level of the body (13). However, Burdett et al. reported the absence of a correlation (r=0.18) when work measures were expressed per unit distance traveled (14). Neither author investigated mechanical measure W_1.

This study addressed the relation between metabolic energy expenditure and mechanical work measures of the prosthetic lower extremity. A significant relation would permit simplified evaluation and feedback in the rehabilitation of the person with amputation. The lack of correlation between W_1 or W_2 and metabolic energy expenditure in subjects with trans-femoral amputation may be related to the fact that these measures were based on the affected limb only. The metabolic demands placed on the person with amputation due to his prosthesis may be reflected in the work performed by his unaffected limb and the remainder of his body.

Hot film anemometers and rate gyroscopes proved to be an effective and automated means of measuring the velocities of the thigh and shank. The precision of hot film velocity measurements is comparable to that of video imaging systems (10), and manual coordinate-tracking is not required. Also, synchronization of film and force plate data is not problematic because all measurements are recorded as electrical signals. Hot films and rate gyroscopes are less expensive than electro-optical systems that automatically digitize data. However, hot film anemometers are fragile, requiring careful handling and a calibrating system. Also, subjects were required to carry an electronics backpack. Rigid mounting of the anemometers and gyroscopes to a prosthesis is easily done, though similar attachment to a normal leg would be more difficult.

The average metabolic energy expenditure in this study for all prosthesis configurations was lower than previously reported (1). The highest energy configurations of this study correspond to the previous reports for traumatic trans-femoral amputees. This may suggest that the measurements of previous studies were made with less energy efficient prostheses.

Table 2.
Summary of average energy data for trans-femoral amputees with various weights at 3 locations (standard deviations are given in parentheses).

Mass (kg)	Energy Measure (Cal/kg/m)	Base Prosthesis		Location of Mass from Knee Axis (cm) 17		25		33	
0	W_1	0.144	(0.063)						
	W_2	0.103	(0.027)						
	Metabolic	1.301	(0.238)						
1.70	W_1			0.152	(0.032)	0.153	(0.033)	0.142	(0.035)
	W_2			0.143	(0.043)	0.158	(0.056)	0.146	(0.026)
	Metabolic			1.108	(0.143)	1.190	(0.191)	1.158	(0.208)
2.84	W_1			0.159	(0.035)	0.158	(0.048)	0.174	(0.063)
	W_2			0.162	(0.046)	0.171	0.047)	0.181	(0.065)
	Metabolic			1.118	(0.237)	1.208	(0.183)	1.238	(0.220)
3.97	W_1			0.158	(0.042)	0.178	(0.056)	0.185	(0.051)
	W_2			0.204	(0.068)	0.201	(0.049)	0.211	(0.077)
	Metabolic			1.160	(0.213)	1.150	(0.198)	1.263	(0.272)

CONCLUSIONS

Added mass, mass location, W_1, and W_2 of the prosthetic extremity were rejected as predictors of metabolic energy expenditure (cal/kg-m) at a significance level of 0.05. Measurement of mechanical work of the prosthetic limb would not be beneficial in managing the rehabilitation of persons with trans-femoral amputation. No relationship was found between W_1 and W_2.

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REFERENCES

1. Bagley AM, Skinner HB. Progress in gait analysis in amputees: a special review. Crit Rev Phys Rehabil Med 1991:3:101–20.
2. Nissen SJ, Newman WP. Factors influencing reintegration to normal living after amputation. Arch Phys Med Rehabil 1992:73:548–51.
3. Waters RL, Perry J, Antonelli D, Hislop H. Energy cost of walking of amputees: the influence of level of amputation. J Bone Joint Surg 1976:58A:42–6.
4. Winter DA. Calculation and interpretation of mechanical energy of movement. In: Hutton RS, ed. Exerc Sport Sci Rev. Philadelphia: Franklin Institute Press, 1979.
5. Aleshinsky SY. An energy 'sources' and 'fractions' approach to the mechanical energy expenditure problem—I. basic concepts, description of the model, analysis of a one-link system movement. J Biomech 1986:19:287–93.
6. Bresler B, Berry FR. Energy and power in the leg during normal level walking. Prosthetic Devices Research Project, University of California, Berkeley, Series 11, Issue 15, 1951.
7. Aleshinsky SY. An energy 'sources' and 'fractions' approach to the mechanical energy expenditure problem—IV. Criticism of the concept of 'energy transfers within and between links'. J Biomech 1986:19:307–9.
8. Schutte LM. Determination of inertial properties of the body for sports biomechanics. Biomechanics in Sport—A 1987 Update 1987:6:27–31, DE-Vol 13, BED Vol 6.
9. Sun SC, Mote CD Jr, Skinner HB. Use of hot wire anemometry to measure velocity of the limb during human movement. J Biomed Eng 1992:14:435–39.

10. Soudan K, Dierckx P. Calculation of derivatives and Fourier coefficients of human motion data, while using spline functions. J Biomech 1979:10:21–6.
11. Durnin JVGA, Passmore R. Energy, work and leisure. Toronto: Heinemann Educational Books Ltd., 1967.
12. Devore JL. Probability & statistics for engineering and the sciences. Monterey: Brooks/Cole Publishing Co., 1982.
13. Ralston HJ, Lukin L. Energy levels of human body segments during level walking. Ergonomics 1969:12:39–46.
14. Burdett RG, Skrinar GS, Simon SR. Comparison of mechanical work and metabolic energy consumption during normal gait. J Orthop Res 1983:1:63–72.

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The effects of thigh soft-tissue stiffness on the control of anterior tibial displacement by functional knee orthoses

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Abstract—Using three soft-tissue analogs of variable compliances, four custom functional knee orthoses were evaluated for their abilities to control anterior tibial displacement (ATD) using an anterior cruciate ligament (ACL)-deficient surrogate knee model with applied forces from 25 to 250 N. These analogs had stiffnesses (compliance^{-1}) ranging from 2.18 N/mm to 4.6 N/mm, simulating the range in the thigh soft-tissue compliances found in subjects ranging from sedentary individuals to competitive athletes. Significant differences in the ATDs allowed were observed between the soft-tissue analogs, orthoses, and the force applied. At low forces, soft-tissue compliance did not play an important role in the reduction of ATD; however, at high forces ATD was directly related to the soft-tissue compliance.

Key words: *anterior tibial displacement, custom functional knee orthoses, soft-tissue compliance.*

INTRODUCTION

Many studies, as well as reviews, have compared functional knee orthoses with respect to resistance to anterior tibial displacement (ATD) in the anterior cruciate ligament (ACL)-deficient knee (1–5). Although the bilateral hinge-post-shell design has repeatedly been shown to provide the greatest degree of resistance to tibial translation (1,5,6), many other factors, including specific design and materials as well as patient physique, affect the efficacy of these orthoses. Soft-tissue composition and compliance of the thigh have been inferred as important factors affecting orthosis resistance to tibial translation by several authors (1,4,5), but the precise effect and degree that this variation in soft tissue has on orthosis function has not been specifically studied. Because thigh tissue compliance varies tremendously between patients with different physiques, ACL-deficient versus ACL-intact, both in terms of tissue composition and muscle conditioning post-injury, it is reasonable to assume that both choice of orthosis and its design will vary depending on which patient population is targeted for treatment. The purpose of this study is to examine the effect of thigh soft-tissue compliance on control of ATD by custom functional knee orthoses.

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METHODS

Orthoses

Four custom functional knee orthoses were evaluated (**Figure 1**):

1. The Performer, Orthopedic Technology, Inc., San Leandro, CA 94577 (Orthotech)
2. Townsend Custom, Townsend Design, Bakersfield, CA 93309 (Townsend)
3. DonJoy Custom 2000, Smith and Nephew DonJoy Inc., Carlsbad, CA 92008-6601 (DonJoy)

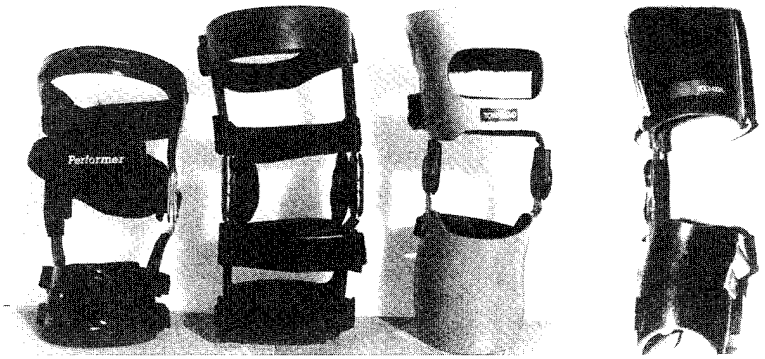

Figure 1.
Four custom functional knee orthoses: from left to right, the Performer, DonJoy, Townsend, Poly-Axial.

4. Poly-Axial Knee Cage, Generation II USA, Inc., Bothell, WA 98011 (Poly-Axial).

All manufacturers advertise that their orthoses resist or restrain ATD in knees with ACL problems. Manufacturers donated their knee orthoses for testing. Fitting of each orthosis strictly followed manufacturer criteria.

Surrogate Knee

Each component of the surrogate leg consisted of a rolled rectangular steel core with a surrounding rigid foam cast of the leg of a 69 in (1.75 m), 150 lb (68 kg), 28-year-old male runner (**Figure 2**). The mechanical rigidity of each knee orthosis was tested in the sagittal plane of the surrogate knee with the femoral component fixed in space and the tibial/ankle component freely moving in that plane (**Figure 3**). The surrogate knee joint was set at 20 degrees of flexion, the position of maximum knee laxity in the sagittal plane (7). The tibial component could freely slide anteriorly relative to the femur at the level of the tibial plateau to a maximum distance of 27 mm. A hinge at the ankle portion allowed application of a posteriorly directed force at the ankle to produce anterior motion of the tibial component. A tensiometer (Dilon Force Gauge, Camarillo, CA) was employed to measure the applied force at the ankle, and a linear displacement scale (Starret, Athol, MA) was used to measure the ATD.

Soft Tissue

Each tissue substitute consisted of three layers of foam glued together with rubber cement. The resulting rectangular foam sheet was 19-mm thick. Duct tape attached each tissue substitute circumferentially around the immobile femoral component of the surrogate knee. The least compliant soft tissue was constructed using a rigid foam-core thigh covered with PPT (Professional Protective Technology, Langer Biomechanicals, Deer Park, NY) and Spenco (Spenco neoprene soft cushion product, Spenco Medical, Waco, TX) to approximate the soft-tissue hardness of a well-conditioned athlete (soft tissue A). Two soft-tissue analog shells (denoted as soft tissues B and C) were constructed using combinations of soft and rigid foam layers.

The compliance (stiffness^{-1}) of the human thigh soft tissue, 7 cm above the knee joint, was measured for 10 (5 female, 5 male) competitive college athletes with an average age of 20 yrs (17–23), 10 (5 female, 5 male) recreational athletes with an average age of 30 yrs (25–35), and 10 (5 female, 5 male) sedentary subjects with an average age of 45 yrs (34–55). The sedentary individuals reported little or no routine athletic activity or exercise. The recreational athletes used in this study are defined as individuals who participate in aerobic exercise for at least one-half hour three times a week. Competitive athletes used in this study participated in college sports. Measurements of the quadriceps were taken during both muscle contraction and relaxation. For the relaxed state, all participants were in a sitting position with the hips and knees in 90° and the feet making contact with the floor. For the contracted state, all participants stood in a semi-squat position with their backs against the wall and their hips and knees at 45°.

The stiffness of the surrogate limb soft-tissue analog and human-subjects' thigh soft tissue was measured with a custom ring transducer in series with a linear variable differential transducer (LVDT) to measure displacement.

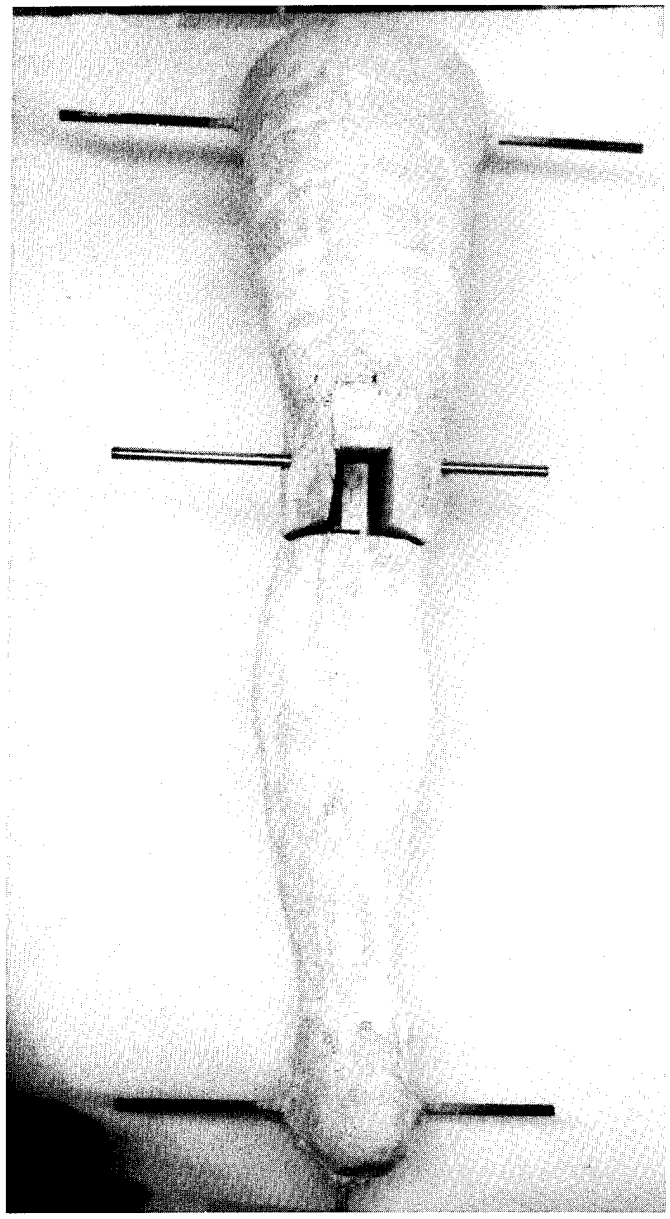

Figure 2.
Surrogate leg.

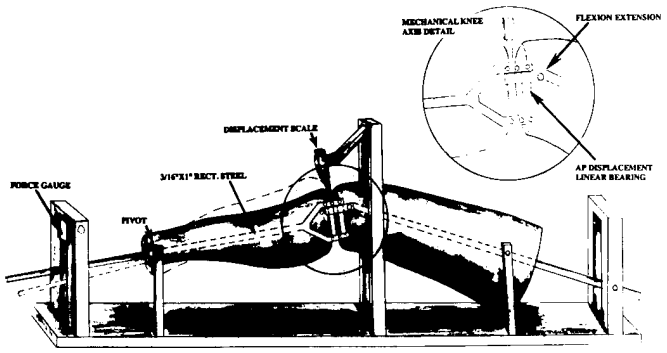

Figure 3.
Surrogate leg with attached force gauge, displacement scale, and a close-up view of the mechanical knee joint.

The compliance of the LVDT was negligible. A 0.25-in (1.22 cm) diameter, flat-faced plunger was used as the indenter in all cases. The housing of the LVDT was held rigid while the force handle drove the LVDT core, with the indenter attached, into the simulated tissue. A maximum force of 20.0 N was used in all cases. The force and displacement data were recorded simultaneously via computer and saved for later analyses. The stiffness of the soft tissue was determined by computing the slopes of the displacement versus applied force plots, using standard linear regression. All relationships between force and displacement were observed to be linear ($R>0.9$). A least squares linear regression was performed to obtain the slope of the force-displacement line, which is equivalent to the stiffness of the material. For the surrogate limb, 4 tissue measurements were made at circumferential locations 7 cm above the knee joint and 5 cm below the joint line at the tibia; the stiffness values were averaged for each location. The same apparatus was also used to measure the compliance of the thigh soft tissue of the volunteers in the relaxed and contracted states.

Experimental Procedure

The soft-tissue analog (A, B, or C) was randomly selected and attached circumferentially to the surrogate knee femur with duct tape. Fitting of each orthosis to the surrogate knee with attached tissue substitute followed manufacturer guidelines. After proper placement, the straps were tensioned to 44.5 N (10 lbs), except for elastic straps which were hand-tightened.

Prior to each test, the surrogate leg was cycled five times through its full range of anterior/posterior tibial motion. The randomly selected orthosis was then applied to the surrogate limb. The application of the orthosis and strap tightening was completed with strict adherence to manufacturer recommendations. Anterior tibial forces ranging from 25 to 200 N were then applied at 25-N increments and the tibial displacement was recorded at each force. Using this sequence, each orthosis was tested 11 times for each soft-tissue analog. After each trial, the orthosis was repositioned on the knee and the straps were retightened as outlined above.

The ATD at each applied force was recorded. When the knee registered its maximum ATD of 27 mm, the trial ended. The entire sequence of 11 trials was repeated for the Orthotech, Townsend, Poly-Axial, and DonJoy orthoses (in random order). Following the testing of all these orthoses, the soft-tissue analog was removed and a different soft tissue was attached around the surrogate knee. The above testing sequence was repeated again for the four orthoses in random order. All three soft-tissue analogs were tested with all four orthoses. ATD values were recorded and compared for all the orthoses, with each soft-tissue analog at all the forces.

Statistical Analysis

For each orthosis, 11 values were obtained for ATD at each force using each tissue substitute. From this data, a mean ATD at each force was calculated. A two-way analysis of variance (ANOVA) by orthosis make and tissue substitute was performed on these means. The results of this two-way ANOVA allowed for a comparison of total mean ATDs using different soft-tissue substitutes. We also performed a one-way ANOVA by orthosis make at 100 N applied force and then performed a multiple pairwise comparison using the Student-Newman-Keuls method. This allowed for a comparison of the mean ATD for each orthosis while keeping tissue conditions constant. Comparisons were made for athletes, recreational athletes, and sedentary subjects in both the relaxed and contracted states. Furthermore, the femoral soft-tissue stiffness of the surrogate limb and the human subjects was compared. All statistical analyses were performed by the statistical computer package, SigmaStat, at a p level of 0.05.

RESULTS

Table 1 shows the average compliance of soft-tissue analogs and the human subjects. The three soft-tissue substitutes are denoted as A, B, and C and range from least to most compliant (A=4.6 N/mm, B=3.26 N/mm, and C=2.18 N/mm; in all cases $p<0.05$). The thigh soft-tissue stiffness of the sedentary individuals averaged 1.43 N/mm (relaxed) and 2.05 N/mm (quadriceps contraction). The recreational athletes showed thigh soft-tissue stiffness of 1.56 N/mm (relaxed) and 2.84 N/mm (contracted). The competitive athletes showed thigh stiffness of 1.71 N/mm (relaxed) and 3.28 N/mm (contracted). There was a significant difference in the contracted compliance of the three groups in the contracted state ($p<0.01$), but not during relaxation ($p=0.59$).

Figures 4 and **5** show the displacement that each orthosis allowed versus the applied forces using each of the three different soft-tissue substitutes. In general, at higher forces the less compliant the soft-tissue analog, the less displacement allowed by the orthosis.

At low forces, the Poly-Axial orthosis (**Figure 4**) showed significant differences in ATD between soft-tissue analogs A, B, and C. At forces higher than 75 N, ATD correlated directly with the compliance of the soft-tissue analogs. At 125 N and above, this orthosis could not be tested further, having reached the apparatus maximum ATD of 27 mm.

For the Performer (**Figure 4**), significant variations in ATD for the three soft-tissue analogs were observed only at forces greater than 125 N. Using the soft tissue A, the least compliant, the Performer showed more ATD than with the more compliant soft tissue B at all forces up to 150 N. Beyond this force, however, soft tissue B allowed full-scale displacement, whereas the soft tissue A did not displace to full scale until forces were greater than 200 N. At no force did soft tissue B allow more displacement than soft tissue C.

The DonJoy orthosis (**Figure 5**) showed no significant difference in displacements between soft tissues B and C until forces of 150 N were applied. At all forces above 150 N, the displacement differences between all three groups were significant to at least $p<0.05$.

The Townsend custom orthosis (**Figure 5**) showed an increase in average tibial displacement with decreasing

Table 1.
Soft-tissue compliance of soft-tisue analogs (a, b, c) and human subjects.

Soft-Tissue Analog (N/mm)	(p<0.05)	
A	4.6 ± 0.28	
B	3.26 ± 0.49	
C	2.18 ± 0.45	
Human Subjects (N/mm)	**Relaxed (p=0.59)**	**Contracted (p<0.01)**
Sedentary	1.43 ± 0.21	2.05 ± 0.37
Recreational Athlete	1.56 ± 0.22	2.84 ± 0.39
Competitive Athlete	1.71 ± 0.12	3.28 ± 0.14

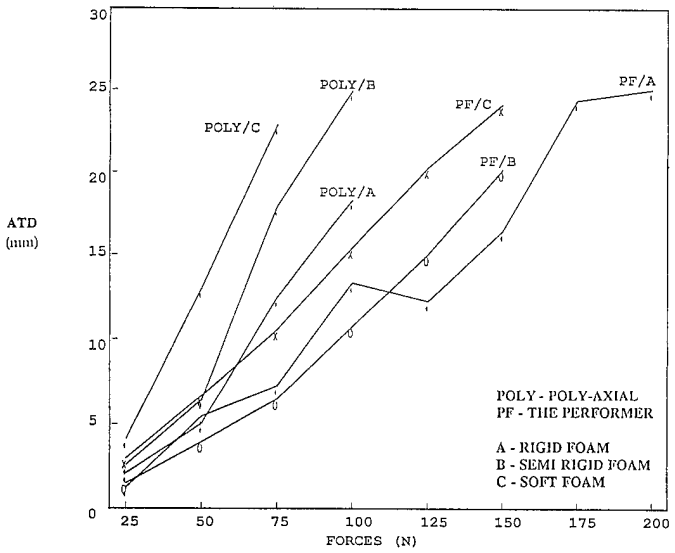

Figure 4.
The role of soft-tissue compliance on functional knee orthosis (Poly-Axial and the Performer) performance.

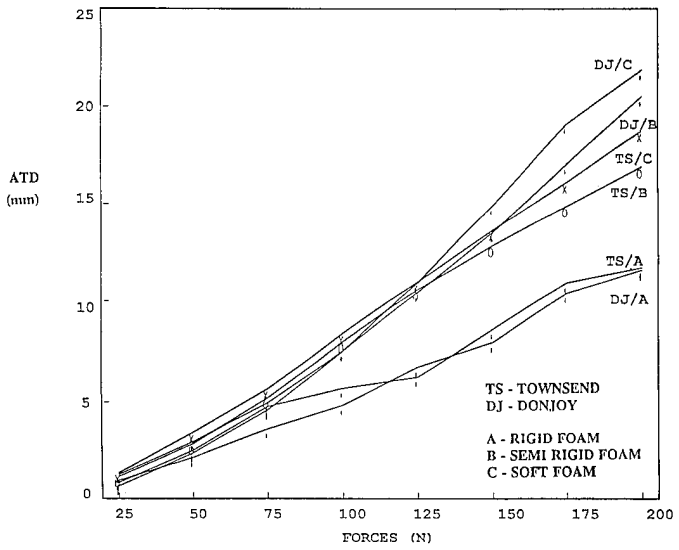

Figure 5.
The role of soft-tissue compliance on functional knee orthosis (Townsend and DonJoy) performance.

compliance at all forces measured. These differences, as seen with the DonJoy orthosis, did not become significant to $p<0.05$ until forces at or above 150 N were applied.

At all forces, the Performer allowed greater ATD than either the DonJoy or the Townsend orthosis. The DonJoy and Townsend orthoses exhibited the best control of ATD and showed no significant statistical difference at both low and high forces. At high forces, with high soft-tissue stiffness, the Townsend orthosis demonstrated the greatest resistance to ATD. The Poly-Axial orthosis, when compared with the other orthoses, reached the 27-mm limit at the lowest force for all soft tissues tested.

DISCUSSION

To our knowledge, no previous studies have investigated the effects of thigh soft-tissue compliance on the control of ATD by custom functional knee orthoses. Reduced ATD was demonstrated by all functional knee orthoses tested. The degree of reduction depended on the compliance of the soft-tissue analog, the type of orthosis, and the force applied.

One major drawback of the surrogate knee used in previous orthosis testing is the lack of quantification of the soft tissue surrounding the knee. The stiffness of each of the soft-tissue analogs used for this study lie within the range of the thigh soft-tissue stiffnesses seen in sedentary, recreational, and well-conditioned athletes. Interestingly, significant differences between the recreational and competitive athletes and sedentary individuals were seen only during quadriceps contraction and not during the relaxed state. It is significant that these differences in thigh soft-tissue compliance exist and depend upon the athletic activity of the individual. These results are especially important today because of the increased use of functional knee orthoses by all segments of the population. As soft-tissue stiffness may indeed be the limiting factor in functional knee orthosis performance (8), this parameter is important in both static bench testing and *in vivo*.

A significant difference in control of ATD was observed when the soft-tissue substitute of the anterior thigh was varied from low to high compliance. Each of the knee orthoses tested demonstrated this dependence on compliance for resistance to ATD. For the Poly-Axial orthosis, the effect of soft-tissue compliance was evident at low forces, but was insignificant when compared to the overall resistance of the orthosis to ATD. On the other hand, for the other three orthoses, varying the soft-tissue compliance affected the resulting ATD, primarily with the application of high forces. Although these three orthoses showed adequate resistance at low forces, the ability to resist ATD was diminished by the effect of soft-tissue com-

pliance when higher forces were applied. In addition, when tested with soft-tissue analogs B or C, none of the orthoses offered enough resistance to limit the amount of displacement to 27 mm with an applied force of over 200 N. Clinical application of this data could indicate that an ACL-deficient knee with bulky and tight thigh musculature (one with less compliance) could facilitate the functional knee orthosis in reducing ATD.

Three of the four orthoses tested in this study (Performer, DonJoy, Townsend) had bilateral hinge-post-shell design that had been shown previously to be the most effective type of orthosis design in restricting ATD (1,5,6). However, these orthoses differ significantly in the degree to which they control ATD. The Poly-Axial orthosis, a unilateral-post-shell design, displayed the greatest amount of ATD at all forces, regardless of the soft-tissue analog. These observations concurred with the results of previous studies (5,6).

ATD is directly related to the force applied. With increased force, all knee orthoses were less effective in controlling ATD. These results correlated with previous studies (1,6,9). Radiographic studies have determined the amount of ATD in subjects with intact ACLs to be 9.8 mm with a range of 5.4–14 mm (10). At low forces, all orthoses were effective in controlling ATD to less than 10 mm. At the highest force applied, 200 N, using the least compliant soft-tissue analog, both the Townsend and the DonJoy orthoses restricted ATD to approximately 10 mm. These observations support the effectiveness of these orthoses in controlling ATD in an ACL-deficient knee at these forces. In addition, the effectiveness of these orthoses depended on the compliance of thigh soft tissue, that is, only low soft-tissue compliance was effective. These observations have not been previously reported.

Because of the tremendous increase in use of functional knee orthoses in patients with ACL deficiencies, as well as those with ACL-reconstructed knees, it is of great importance that the factors of patient physique which affect orthosis efficacy be determined and controlled for as well as possible. The results of this study show that these factors can effect control of ATD by knee orthoses at forces as low as 150 N, forces below those generated during strenuous activities. Thus, in order to employ the functional knee orthosis most effectively, rehabilitation to improve quadriceps strength and bulk will be important not only for older patients with less muscle mass, but also for patients with quadriceps atrophy following surgery as well as for deconditioned athletes. Physicians should consider this information when prescribing a custom functional knee orthosis to an ACL-deficient or an ACL-reconstructed patient who plans to engage in strenuous activity.

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REFERENCES

1. Beck C, Drez D Jr, Young J, et al. Instrumented testing of functional knee braces. Am J Sports Med 1986:14:253–6.
2. Butler PB, Evans GA, Rose GK, et al. A review of selective orthoses. Br J Rheumat 1983:22:109–15.
3. Cawley PW, France EP, Paulos LE. Comparison of rehabilitative knee braces: a biomechanical investigation. Am J Sports Med 1989:17:141–6.
4. Cawley PW, France EP, Paulos LE. The current state of functional knee bracing research: a review of the literature. Am J Sports Med 1991:19:226–33.
5. France EP, Cawley PW, Paulos LE. Choosing functional knee braces. Clin Sports Med 1990:9:743–50.
6. Liu SH, Lunsford T, Vangsness CT, et al. Comparison of functional knee braces for control of anterior tibial displacement. Clin Orthop 1994:302:147–50.
7. Markolf, KL, Mensch JS, Amstutz HC. Stiffness and laxity of the knee—contributions of the supporting structures. J Bone Joint Surg Am 1976:58(A):583–7.
8. Beynnon BD, Pope MH, Wertheimer CM. The effects of functional knee braces on strain on the anterior cruciate ligament in vivo. J Bone Joint Surg Am 1992:9:1298–312.
9. Branch TP, Hunter R, Reynolds P. Controlling anterior tibial displacement under static load—a comparison of two braces. Orthopedics 1988:11:1249–52.
10. Granberry WM, Noble PC, Woods GW. Evaluation of an electrogoniometric instrument for measurement of laxity of the knee. J Bone Joint Surg Am 1990:72(A):1316–22.

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Volumetric determinations with CAD/CAM in prosthetics and orthotics: Errors of measurement

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Abstract—During the next decade CAD/CAM technique will probably become routine in prosthetics and orthotics, not only as a complement to manual techniques, but also introducing new possibilities. However, even complex and sophisticated techniques have errors of measurement that must be considered. Such errors are of two principal kinds: systematic errors and random errors. In this study we have evaluated the Swedish CAPOD system with respect to volumetric determinations. We used two types of reference objects for volume determinations: cylinders and amputation residual limb models. Three different sizes were examined of each type of object. Volume measurements with CAPOD were compared with volumes obtained by water immersion or mathematical calculation (cylinders only). We found a constant, linear systematic error of +2.5%. Such an error can easily be corrected for. The random error, represented by the coefficient of variation, was 0.5%, which means that there is a theoretical possibility to detect volume changes exceeding 1%. We consider the precision sufficient for clinical practice in prosthetics and orthotics. Biological variations due to soft tissue deformation must be added on top of these errors. Such deformations were not evaluated in this study.

Key words: *amputation, CAD/CAM, errors of measurement, trans-tibial, volumetric.*

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INTRODUCTION

During the last decade, several CAD/CAM systems have been developed for use in prosthetics and orthotics (1,2). The potential advantages of this new technique are 1) even quality of prosthetic sockets, 2) time saving, 3) better fitting of sockets, and 4) lower cost. The new technique can be used not only as a complement to traditional prosthesis making, but also in new applications. For example, volumetric determinations can be used to decide the proper time for permanent prosthetic fitting. The computer technique also makes computer integrated manufacturing (CIM) possible, integrating the production of prosthetic sockets with administrative routines such as patient records, appointments, deadlines for follow-up, and economy (3). Nevertheless, use of the CAD/CAM technique is rare in daily prosthetic and orthotic practice, and is still considered exclusive. Over the next decade, however, with further technical advances, we can foresee an increased use of this technique, which will probably become routine (3). With more sophisticated and complex instruments, there is a danger that we rely too much on the instruments and do not consider the errors of measurement. All instruments, no matter how sophisticated, have errors of measurement. Error estimations related to calibration methods for CAD/CAM have been published by Kofman et al. (4), but to our knowledge no study has yet been published on measurement errors in CAD/CAM technique applied to residual limb models or real residual limbs.

In all kinds of experimental work, the degree of accuracy and precision is of utmost importance. Errors of measurement are unavoidable, and they can—especially if they are unknown—seriously jeopardize the whole measurement. If, however, they are known and quantified, it is possible to handle them and maybe also to correct for them. If we do not know anything about the measurement errors, the measurement will be of almost no value. For this reason it is necessary to estimate such errors, especially if the instrument is to be used in research studies.

Accuracy and precision are not ends in themselves. In basic research we do not always know the final use of a measurement or a new technique. Consequently, it may be justified to be more careful than what is demanded from the actual problem, because the extra accuracy and precision may be needed to solve specific problems later on. Furthermore, the more accurate and precise, the more fields of application the new technique will find.

The sources of measurement errors may be (5,6):

- the observer/experimenter
- the instrument
- the environment
- the object under observation.

The experimenter may be tired or careless, or poorly informed or trained. There may be no standardized measuring procedures or guidelines for the measurement. The instruments may be badly calibrated, the algorithms for calculation may be erroneous, there may be approximation errors, the environment may not be standardized, the test objects may change during the measurement, etc.

Three questions of interest may arise in the context of measurement (7): 1) How accurate are the measurements—are we measuring what we are supposed to be measuring (accuracy, validity)? 2) How reproducible are the numbers we produce (precision, reproducibility, reliability)? and 3) What do those numbers convey about the condition of the test patient (utility)? The first two aspects are considered in this study.

Error Calculation

A test measurement x_i (x_i denotes an arbitrary measurement value) is in itself quite worthless if it is not related to the "true" value ξ_i and the error of measurement ε_i. We can assume the following simple structure for a test measurement:

$$x_i = \xi_i + \varepsilon_i \qquad [1]$$

Errors of measurement are of two main types (8,9): systematic errors (bias, deterministic errors, inaccuracy) and random errors (irregular errors, stochastic variation, imprecision). If we know the "true" value, the systematic error or bias is the difference between the average of a number of test measurements and the "true" value. This error may be constant or it may follow some mathematical function, for example a periodic error. It can be impossible to separate instrument errors from errors due to the experimenter.

There is always a true value, but often we do not know it. If, however, measurements can be performed a number of times on a standardized test object, under strictly standardized conditions, and with methods that are well established, the average values from such measurements can be used as reliable estimates of the "gold standard." The best estimate of the gold standard is the mean of repeated measurements with calculation of the boundaries within which it may be included (8,10). In practice, the systematic error of a measurement is essentially a comparative judgment—a measure of the extent to which the average of different measurements agree or disagree in terms of results obtained by measurement on a given test object. We have used this method in the present study.

According to the central limit theorem, random errors follow the Gaussian distribution (7,9):

$$\varphi(\varepsilon) = \frac{1}{\sigma\sqrt{2\pi}} e^{\varepsilon^2/2\sigma^2} \qquad [2]$$

Given a sample of n replicated measurements x_i on a single test object, the sample mean μ should converge in probability (as n increases) to the "true" value ξ_i, unless there is a systematic error (bias). The error mean μ_ε converges to zero, and the error variance σ^2_ε is equal to the variance σ^2 of the replicated measurements. The standard deviation σ is a direct measure of the imprecision (random error) of the measurement (8).

Often the error is calculated as the relative error, that is, ε_i/ξ_i. ξ_i is unknown. If, however, the error is small, the relative error can be calculated as ε_i/x_i or $100 \times \varepsilon_i/x_i$ (percentage error relative to the measurement). The error of measurement can never be estimated by one single measurement. To reveal errors of measurement, the same observer can perform repeated measurements on the same test object.

If the measured data are to be used in further calculations, the problem of error propagation must be considered. Such error propagation can be calculated with

methods based on differential calculus, but are not considered in this article (6,9–11).

Aim of the Study

If CAD/CAM technique is used in research and practice, it is necessary to evaluate the different kinds of measurement errors associated with this technique. In a previous study, we have compared CAD/CAM made prosthetic sockets with conventionally made ones (12). We used the Swedish CAPOD system (13). The aim of the present study was to examine errors of measurement of the CAPOD system in volumetric measurements using cylinder and residual limb models with known volumes.

METHOD

Test Objects

Two types of objects were used: cylinders and residual limb models of different sizes.

Cylinder models: Three homogeneous aluminum models were used. The diameters were 69.5 mm, 100.0 mm and 150.0 mm respectively. The height of the models was 200.0 mm. They were measured with a vernier caliper and their surfaces were painted white to increase their reflectivity to laser light.

Residual limb models: Three plaster-of-Paris models of below-knee amputation residual limbs were made. Three different sizes were used. They were lacquered several times to make them water resistant.

Reference Measurements of Volume

Water immersion technique: The test objects were immersed in a vessel filled with de-ionized water. At the water level there was a spout, through which any displaced water was released. The cylinders were totally immersed, and the residual limb models were partially immersed to a black level mark at the position of the knee joint. The displaced water was collected and weighed on a precision balance. Temperature and atmospheric pressure were recorded in order to have control over considerable alterations. The density of de-ionized water at 22° C is 0.997770 kg/dm^3. From the weight of the displaced water, the volume of the immersed object was calculated. Five repeated measurements were performed on each object. The experimental set-up is shown in **Figure 1.**

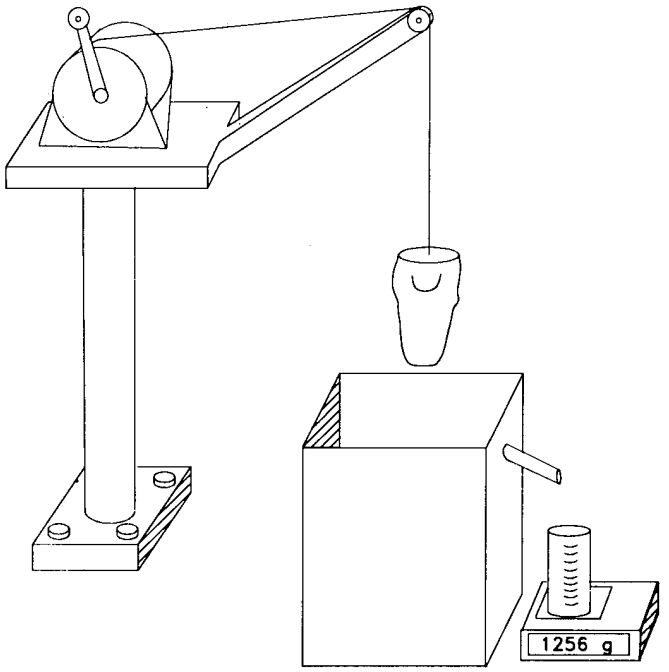

Figure 1.
Equipment for volume measurement with water immersion technique.

Due to the capillary forces (forces of surface tension and adhesive forces acting on the test object) the water surface is curved near the test object. This phenomenon could introduce a small error of measurement. This error was calculated and was found to be less than 0.1 percent (the calculations are not included in this article). We regard this error small enough to be discarded in the further calculations.

Mathematical calculation of cylinder volumes: The cylinder volumes were calculated according to standard mathematics.

CAPOD measurements of volume: The CAPOD system has been described in an earlier article by Öberg et al. (13). The system consists of a laser scanner for noncontact scanning, a CAD software to perform socket design rectifications and a milling machine to mill the socket mold. The CAD/CAM system was used without any modifications for the study. The system was calibrated according to the guidelines given by the CAPOD Systems CO (4). Two different laser scanners were tested: one in Jönköping and one in Uppsala, Sweden.

The test objects were scanned, and the volumes were computed. For the cylinders, the total volume was calcu-

lated; for the residual limb models, the volume up to the black level mark was calculated. Five repeated measurements were performed on each object.

Statistical Analysis

Means, variances, standard deviations (SD), coefficients of variation (CV, standard deviation as percent of the mean value), and students *t*-values were calculated according to standard procedures (14).

RESULTS

Systematic Errors

The absolute volumes of the cylinder and residual limb models, calculated with different methods, are shown in **Tables 1** and **2.** In **Tables 3** and **4**, the differences between the CAPOD measurements and the results obtained with the reference methods are shown.

Cylinder models: For the cylinder models, the differences are not consistent and in many cases not statistically significant. **Figure 2** shows that the measurements with all methods are close to each other.

Residual limb models: For the residual limb models there was a systematic difference of about 2.5 percent in all cases. The CAPOD system overestimated the volumes. The differences were statistically significant and of the same magnitude for both suites of test equipment. These differences are clearly seen in **Figure 3.**

Random Errors

The random errors were estimated by SD and CV. The values are shown in **Tables 1** and **2.** The CV was very small, in most cases below 0.5 percent. Thus, 95 percent of the random variation may be included within an interval of mean volume ±1 percent (mean ±2CV).

DISCUSSION

One idea underlying the introduction of CAD/CAM technique in prosthetics and orthotics has been that contacting methods may introduce measurement errors due to deformation of the soft tissues or plaster cast, while non-contacting optical scanning may enable measurement of higher accuracy (4). But CAD/CAM methods cannot only replace conventional methods, they also have a new potential of themselves. For example, volume determination can be used to decide when postoperative volume changes cease. This is probably the proper time for permanent prosthetic fitting. Hitherto, we have not had simple methods for such volume determinations. Fernie et al. (15) developed one complex method to evaluate volume fluctuations and some other volume studies with water displacement methods have been performed (16,17).

We chose two different methods for volume determinations as references to represent the gold standard. The mathematical calculations have no systematic errors, and the random errors were considered negligible. The water

Table 1.
Cylinder volumes (ml). Means, standard deviations (SD) and coefficients of variation (CV).

Method		Model 1	Model 2	Model 3
Mathematical calculation		758.7	1572.4	3534.3
Water immersion	mean	754.9	1570.1	3529.3
	SD	3.04	13.30	7.35
	CV	0.40%	0.85%	0.21%
CAPOD Jönköping	mean	770.0	1583.0	3518.8
	SD	5.39	4.06	1.9
	CV	0.70%	0.26%	0.05%
CAPOD Uppsala	mean	768.2	1566.0	3523.6
	SD	4.82	7.65	9.10
	CV	0.63%	0.49%	0.26%

Table 2.
Stump volumes (ml): Means, standard deviations (SD) and coefficients of variation (CV).

Method		Model 1	Model 2	Model 3
Water immersion	mean	1020.00	1202.2	1458.6
	SD	6.71	6.41	15.76
	CV	0.66%	0.53%	1.08%
CAPOD Jönköping	mean	1051.0	1234.0	1489.8
	SD	6.44	4.53	8.14
	CV	0.61%	0.37%	0.55%
CAPOD Uppsala	mean	1045.2	1233.4	1492.8
	SD	3.56	3.13	5.76
	CV	0.34%	0.25%	0.39%

Table 3.
Difference between mean volumes (ml): Cylinder models.

Method		Model 1	Model 2	Model 3
CAPOD-J	Mathematical calculation	11.3	10.6	−15.5
		(1.49%)	(0.67%)	(−0.44%)
		t=4.90	t=6.04	t=−18.02
		p<0.01	p<0.01	p<0.001
CAPOD-U	Mathematical calculation	9.5	−6.4	−10.7
		(1.25%)	(−0.41%)	(−0.30%)
		t=4.64	t=−1.87	t=−0.42
		p<0.01	N.S.	N.S.
CAPOD-J	Water immersion	15.1	12.9	−10.5
		(1.99%)	(1.07%)	(−0.72%)
		t=3.39	t=3.67	t=−1.32
		p<0.01	p<0.01	N.S.
CAPOD-U	Water immersion	13.3	−4.1	−5.7
		(1.73%)	(0.26%)	(−0.16%)
		t=3.30	t=−0.38	t=−0.76
		p<0.05	N.S.	N.S.

N.S. = Not significant J = Jönköping equipment U = Uppsala equipment

Table 4.
Difference between mean volumes (ml): Stump models.

Method		Model 1	Model 2	Model 3
CAPOD-J	Water immersion	31.0	31.8	31.2
		(3.04%)	(2.65%)	(2.14%)
		t=7.45	t=9.07	t=3.93
		p<0.001	p<0.001	p<0.01
CAPOD-U	Water immersion	25.2	31.2	34.2
		(2.47%)	(2.53%)	(2.29%)
		t=7.42	t=9.39	t=4.56
		p<0.001	p<0.001	p<0.01

J = Jönköping equipment U = Uppsala equipment

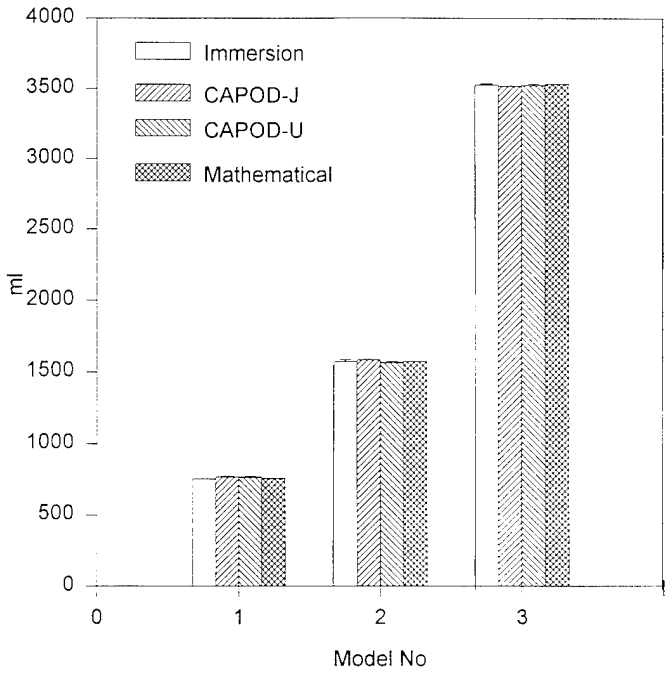

Figure 2.
Diagram showing mean cylinder model volume, measured with four different methods. Error bars indicate the standard deviation.

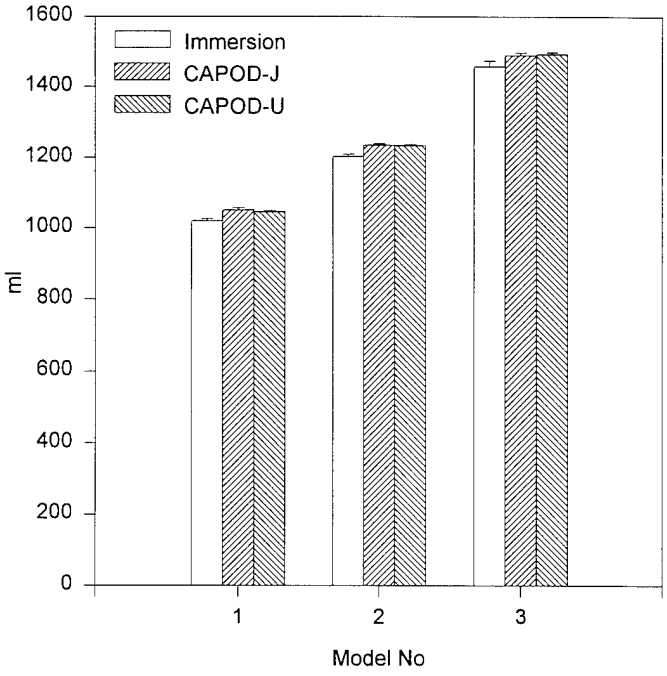

Figure 3.
Diagram showing mean residual limb model volume, measured with three different methods. Error bars indicate the standard deviation.

immersion method is well established since the days of Archimedes, and it was considered free from systematic errors. Thus, the mean of repeated measurements could represent the "true" volume of both cylinders and residual limb models.

For the cylinder models, we found a statistically significant difference between CAPOD measurements and the reference methods only for the smallest model. For the residual limb models, however, we found a systematic error of about +2.5 percent for all models. These results

indicate that the calculation algorithm of the CAPOD system gives a good approximation of the volume for simple, regular objects such as a cylinder, but systematically overestimates the volume of irregular objects such as amputation residual limbs. The difference in the results can be due to methodological differences between measuring the residual limb model and the cylinder model. The error, however, was linear and almost constant throughout the range of measurement, and can easily be corrected for. The CAPOD system has an option for scaling up and down the dimensions of the scanned object.

The reproducibility was good, as represented by the CV, and was less than 0.5 percent (i.e., less than the variation of the water displacement method). Consequently 95 percent of random errors will be included within an interval of ±1 percent. Kofman et al. (4) present two calibration and 3-D reconstruction methods, dependent and independent of camera parameters. They found that their measurements were more than adequate for residual limb-shape measurement and prosthetic fitting. In their study, however, calibration measurements were made on a stepped cylinder with known dimensions. No measurements were performed on a residual limb model or a real residual limb. Thus it is only possible, from their study, to draw conclusions about their test object, not about residual limb models or residual limbs, as measured in the present study.

We want the measurement errors to be as small as possible, but more accuracy and more precision requires greater complexity and cost. If, as was the case in this study, the systematical error is easy to correct for, this should be done, otherwise we must consider whether the error is acceptable or not. The interpretation of the random errors depends on the tolerance and precision that is necessary. Volume changes larger than 1 percent can be detected with the CAPOD system.

Volume determinations are also important as validity tests of the CAD/CAM technique. As one of the ideas was a reduction of errors, it is important that the new technology does not introduce new errors that are larger than the old technique.

In the clinical situation we do not measure standard objects, but living people with residual limbs consisting of soft tissues that can be deformed. Such variations are not related to the CAD/CAM system per se, but must be added on top of the other random errors. Such errors are, however, not the topic of this study.

It is important for users of different measuring devices to control accuracy and precision. This can be done by regular and standardized calibration measurements. The equipment must be checked with reference objects of known form and volume. It is also important to use reference objects that simulate real test objects, such as amputation residual limbs. Such test objects should follow the delivery of CAD/CAM systems, and the calibration procedures must be simple and preferably automatic.

CONCLUSIONS

In the present study, we have examined the accuracy (validity) and precision (reproducibility) of a CAD/CAM system—the Swedish CAPOD system—when used for volumetric determinations. We found a linear, almost constant systematic error of +2.5 percent, which could easily be corrected for, and a small random error, represented by a CV of less than 0.5 percent.

It is important that new technical systems are evaluated with respect to errors of measurement, not only by the manufacturer, but also by independent research groups. To our knowledge, the present study of the CAPOD system is the only one of its kind. We believe that unknown errors of measurement are even more dangerous than known errors, because the latter can often be corrected or compensated for. In our opinion, the systematic error should be corrected for by the built-in possibility to scale the volume up and down. The precision permits detection of changes larger than 1 percent. We consider the precision sufficient for routine clinical practice in prosthetics and orthotics.

REFERENCES

1. Brüssel A. CAD/CAM-technologien in der orthopädietechnik: eine kritische analyse. Orthopädie Technik 1990:41:634–42.
2. Brüssel A. CAD/CAM in prosthetik und orthotik—stand der technik. Orthopädie Technik 1991:42:487–95.
3. Lilja M, Öberg T, Karsznia A. The applications of CAD/CAM technique in prosthetics and orthotics. Biomech Sem 7, 1993:157–62. ISSN 1100-2247.
4. Kofman J, Öberg K, Karlsson A, Lindström B. Comparison of camera-parameter dependent and independent methods of optical scanner calibration and 3-D residual limb-shape reconstruction. In: Proceedings of 7th World Congress of ISPO, 1992:152.
5. Ackoff R. Scientific method: optimizing applied research decisions. New York: Wiley, 1970.
6. Bevington PR, Robinson DK. Data reduction and error analysis for the physical sciences. 2nd ed. New York: McGraw-Hill, 1992:1–51.

7. Strike PW. Statistical methods in laboratory medicine. Oxford: Butterworth-Heinemann, 1991.
8. Barford NC. Experimental measurements: precision, error and truth. 2nd ed. Chichester: Wiley, 1985:85–117.
9. Gellert W, Gottwald S, Hellwich M, Kästner H, Küstner H. Calculus of errors, adjustment of data, approximation theory. In: Gellert, et al. The VNR concise encyclopaedia of mathematics. 2nd ed. New York: Van Nostrand Reinhold, 1989:607–13.
10. Topping J. Errors of observation and their treatment. New York: Chapman & Hall, 1974.
11. Deming WE. Statistical adjustment of data. New York: Dover Publications (Facsimil 1964), 1943:37–48.
12. Öberg T, Lilja M, Johansson T, Karsznia A. Clinical evaluation of below-knee prostesis sockets: a comparison between CAD/CAM made and conventionally made sockets. Prosthet Orthot Int 1993:17:164–71.
13. Öberg K, Kofman J, Karlsson A, Lindström B, Sigblad G. The CAPOD system—a Scandinavian CAD/CAM system for prosthetic sockets. J Prosthet Orthot 1989:1:139–48.
14. Armitage P, Berry G. Statistical methods in medical research. 2nd ed. Oxford: Blackwell, 1987.
15. Fernie GR, Holliday PJ, Lobb RJ. An instrument for monitoring residual limb oedema and shrinkage in amputees. Prosthet Orthot Int 1978:2:69–72.
16. Fernie GR, Holliday PJ. Volume fluctuations in the residual limbs of lower limb amputees. Arch Phys Med Rehabil 1982:63:162–5.
17. Golbranson FL, Wirta RW, Kuncir EJ, Lieber RL, Oishi C. Volume changes occuring in postoperative below-knee residual limbs. J Rehabil Res Dev 1988:25:11–8.

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An animal model and computer-controlled surface pressure delivery system for the production of pressure ulcers

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Abstract—Pressure ulcers continue to be a major health care problem. This paper describes an animal model and surface pressure delivery system for the production of experimentally derived pressure ulcers. A method for inducing dermal pressure lesions on the fuzzy rat was developed using a computer-controlled displacement column which produced a constant tissue interface pressure. The pressure column consists of a force transducer located between two 0.5-in (1.27-cm) diameter metal cylinders. The desired cutaneous pressure is maintained by a computer-controlled miniature stepper motor which displaces the column with the aid of interactive software. The force transducer signal is converted from analog to digital form, amplified, and recorded. Blood perfusion is monitored using a laser Doppler flowmeter (located in the tip of the column) during the application of pressure. The application of 145 mmHg pressure for 5 consecutive 6-hr sessions resulted in a greater than 90% incidence of pressure ulcers. The implications of our model and contributions of earlier animal models are discussed. This model provides a tightly controlled and measured environment making possible the scientific study of ulcer development and the evaluation of potential preventative or curative compounds.

Key words: *animal model, decubitus ulcers, fuzzy rat, pressure ulcers.*

INTRODUCTION

Pressure ulcers (bed sores, decubitus ulcers) continue to be a major health care problem affecting a large segment of the population which includes individuals with spinal cord injury, patients with neurological disease, those with immobility, and the aged (1).

A multitude of etiologic, pathomechanical, and pathophysiologic mechanisms are associated with the development of pressure ulcers, including pre-existing neuropathology, immobility, shear, friction, malnutrition, maceration, ischemia, and pressure (1). Unrelieved pressure is generally considered the most important of these factors (1). However, the amount of pressure and critical threshold of reduced blood flow leading to ischemia and pressure ulcer formation is unknown (2). In 1930, Landis (3), using micro-injection to study capillary blood pressure, found an average pressure of 32 mm Hg in the arteriolar limb, 22 mm Hg in the mid-capillary bed and 12 mm Hg on the venous side. A number of authors using various

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methods of measurement have documented pressures at specific anatomical locations at risk for pressure ulcers in humans (4–8). The widely held perception that pressure at the surface of the skin or interface pressure, in excess of 32 mm Hg will lead to closure of capillary beds and result in tissue ischemia remains unestablished in laboratory studies (2). Many clinicians have assumed that pressures below 32 mm Hg are "safe" (2). Although pressure is known to be a primary risk factor for the development of pressure ulcers, the relationship between the magnitude of pressure, the duration of pressure, and tissue perfusion in the development of pressure ulcers is unknown. Evaluation of the generality of the possible link between pressure and tissue perfusion is critical, especially considering that the susceptibility of tissue and muscle varies according to the tissue characteristics at the particular anatomical location (9,10). Additionally, tissue resiliency may be influenced by other characteristics of the patient such as age, diet, and various immunological factors (11). However, pressure as a primary risk factor for the development of pressure ulcers requires more intense study. The following critical questions are relevant to clinical practice and gaps in our research knowledge base: 1) how much pressure, and over what period of time, typically results in ulceration, and 2) what tissues are at greatest risk (2). Because of multiple variables found in the human environment, pressure ulcer research is extremely difficult to conduct in the clinical setting (2). To answer these questions it is necessary to develop an animal model in which variables that contribute to the development of pressure ulcers can be monitored and quantified.

Animal models have been used to study the etiology of pressure ulcers. Groth (12) applied varying degrees of pressure to the posterior ischii of rabbits for varying periods of time. He reached several conclusions: 1) that pressure ulcers simulating those in humans can be produced experimentally; 2) that the larger the muscle mass, the greater the ability to withstand pressure; and 3) that the effective force may be greater below the surface, especially at sites overlying a bony prominence, which might account for the greater destruction of tissue at the base of the inverted cone so often seen clinically. In 1953, Husain (13) reported microscopic changes in rat muscle subjected to a pressure of 100 mm Hg for as little as 1 hour. Husain emphasized the importance of time in the development of tissue anoxia and demonstrated that relatively light pressure over an extended period of time causes more intense tissue necrosis than higher pressure over a shorter period of time. His findings were similar to those reported by Groth (12) and Kosiak (5). Kosiak conducted a classic experimental pressure ulcer study in dogs, which focused primarily on the epidermis. Pressure ranged from 60 to 550 mm Hg for periods of from 1 to 12 hours. Microscopic examination of tissue after 1 hour of pressure application (60 mm Hg) showed significant histological lesions, including inflammatory cell infiltration, extravasation, and hyaline degeneration. Tissues subjected to higher pressures for longer periods of time also showed significant lesions including muscular degeneration and venous thrombosis. In either case, tissue ischemia resulted in irreversible cellular changes and ultimately in necrosis and ulceration. He concluded that low pressure of longer duration was as injurious to tissues as intense pressure of short duration, and that prolonged pressure was the direct and primary cause of pressure ulcers. Daniel, et al. (14) investigated the effect of spinal cord transection on pressure ulcers in paraplegic pigs. Pressure was applied by an electromechanical pressure applicator controlled by a computer. Following tissue atrophy, pressure was applied to the paraplegic animals for varying durations of time. This experimental model produced pressure ulcers extending down to the bone. The resulting tissue damage was assessed and graded, and a pressure-duration curve for paraplegic animals was established. Daniel (14) agreed with Harman (15) and others (16,17) that muscle is extremely sensitive to ischemic damage and that degeneration begins in the muscle as early as 4 hours after the application of pressure. He did not embrace the theory that pressure-induced ischemia was a primary cause of pressure ulcers, but did suggest that an important cause was the inability of the tissue to respond to external loads because of tissue wasting associated with paraplegia, repeated trauma, and infections. Nola and Vistnes (16) documented significant areas of muscle necrosis in rats when pressure was applied to a transposed muscle flap over bone. The model used inverted plastic syringes, driven by compressed air, as pistons to deliver force to the greater trochanter. Force was measured using a pressure transducer system. Dinsdale (18) analyzed the role of pressure and friction in the production of pressure ulcers in normal and paralyzed pigs by utilizing air-driven pistons. Pressure of 160–1120 mm Hg was mechanically applied with and without friction for 3 hrs. The resulting ulcerations extended into the dermis and were present after 24 hrs. Dinsdale concluded that friction contributed to the pathogenesis of pressure ulcers, and he cautioned against blindly accepting the pressure-ischemia relationship as the only cause for ulceration. He also concluded that, al-

though constant pressure applied continuously over a given time period caused irreversible tissue damage, only minimal damage resulted if pressure was intermittently relieved. Ferguson-Pell, et al. recently described a system designed to measure hyperemia in the clinical setting (19, 20) which shares many of the characteristics of our pressure delivery system. Their system is a complex pneumatic skin indentation system suitable for short-term clinical studies. "Blood content" and "blood oxygenation" in control patients and patients with spinal cord injuries were measured using a tissue reflectance spectrophotometer. The authors concluded that the reactive hyperemia response was not substantially different between the two groups of patients.

Contributions and Critique of Previous Models

Although our particular approach is relatively new, we obtained considerable guidance from preceding models. Groth (12), as early as 1942, utilized a balanced beam device to apply pressure to the posterior ischii of rabbits for varying periods of time and pressure via two 15 mm circular discs. Force was applied perpendicular to the skin surface to study pressure ulcer etiology. Skin breakdown did not occur in this model and the applied pressure was not continuously monitored.

In 1953, Husain (13) studied pressure effects by means of a plethysmograph or pressure cuff and reported on microscopic changes in rat muscle. In 1959, a classic study by Kosiak (5) subjected 16 dogs to accurately controlled pressure created by inverted air driven piston syringes. He demonstrated that the magnitude of pressure is an important determining factor in the development of pressure ulcers. This relationship to pressure may be the result of ischemia with respect to the capillary closing pressure, but these variables were not directly assessed. As with the Husain study, the pressure delivery using air-driven syringe pistons had the shortcoming of "metastable" pressure delivery, and no determination of the actual cutaneous pressure.

Although in some cases the pig may be as appropriate as the rat for the study of pressure ulcers, their added expense is prohibitive, and they are not suited for large experimental studies. A major reason we selected the rat as our animal model of choice is that more is known about the pharmacological effects on absorption, distribution, and metabolism of drugs in rats than in any other species.

The recently developed bellows indentation system of Ferguson-Pell et al. (20) was designed to study short-term interventions such as hyperemia in the clinical environment; however, because of its large size this system is not readily adaptable for use in a study of small animals.

This paper describes a new model that incorporates many features for studying the determinants of and potential treatments for experimentally derived pressure ulcers. Although the studies described above provide systems with varying degrees of appropriateness for pressure ulcer study, none of these systems combine the features of reliability, servo control of pressure delivery, simultaneous monitoring of physiological endpoints, cost effectiveness, compactness, accuracy, flexibility, and the ability to handle large numbers of animals simultaneously, all of which are inherent in our system. We describe here a rodent model and pressure delivery system that results in the reliable production of experimental pressure ulcers in a controlled environment.

METHODS

Subjects

Male or female fuzzy rats (21) (Harlan Sprague-Dawley, Inc., Indianapolis, IN), each weighing 150–350 g, can be used to characterize pressure-induced damage to cutaneum and subcutaneous tissue. In the experiments reported here we used rats weighing 200–300 g. Hypotrichotic fuzzy rats were selected to minimize the effect of hair on skin and subcutaneous tissue pressure and to eliminate the need for artificially preparing the skin by shaving or by other methods that could introduce artifact. Rats were housed in groups in a cycle of 12 hrs of light and 12 hrs of dark; food and water were available *ad libitum*. During the series of daily experiments described below, the rats were individually housed under the same conditions—under Institutional Animal Care and Use Committee (IACUC) protocol 90-0051M.

Anesthesia

Each rat was anesthetized by intraperitoneal (i.p.) delivery of a xylazine and ketamine combination at a dose of 14.41 mg/kg Rompun (Miles, Inc., Shawnee Mission, KS) combined with 79.95 mg/kg Ketaset (Fort Dodge Laboratories, Inc., Fort Dodge, IA). At appropriate intervals, supplemental doses were administered to maintain a stable plane of anesthesia over the 6-hr pressure session. At the end of the pressure session, each animal received an i.p. injection of physiologic 2.5 percent dextrose and 0.45 percent saline to maintain adequate hydration.

Procedure

The integration of the various components of the cutaneous pressure delivery system is illustrated in **Figure 1a.** After induction of anesthesia, the animal was placed in a custom-made saddle and restraining device. The base of this device is constructed of durable 9-lb Ethafoam (Dupont, Foam Design, Inc., Lexington, KY) designed to ideally expose the hip areas for delivery of pressure to the skin. Curved Plexiglas braces stabilize the animal when pressure is applied by the pressure columns. During each experiment, the core body temperature of the rat was monitored by a thermistor probe (model 44033, Yellow Springs Instruments, Yellow Springs, OH) inserted rectally, and maintained at 35–36° C by a heat blanket (model T200, Gaymar T Pump, Orchard Park, NY) incorporated into the walls of the restraining device. The base is attached to an adjustable track that provides flexibility when the pressure column is positioned so that the contact surface of the column is parallel to the skin surface over the greater trochanter. This positioning of the column is accomplished by a swivel joint that moves in three dimensions (see **Figure 1b**). When the column is properly positioned over the skin surface, the positioning swivel joint is tightened. After the first pressure session was completed, the pressure indentations on the skin were marked with an indelible marker to insure proper placement of the pressure column in subsequent pressure sessions. During each daily session, pressure is applied for 6 hrs at a pressure of 145 mm Hg, the components of which are 250 g force over the 13 mm diameter surface area at the end of the pressure column. Pressure was measured at the skin surface and applied over the trochanter region. Unless indicated otherwise, pressure was applied for 5 consecutive daily sessions of 6 hrs duration each. This magnitude of pressure was selected on the basis of a review of the pressure ulcer literature and on the results of preliminary experiments.

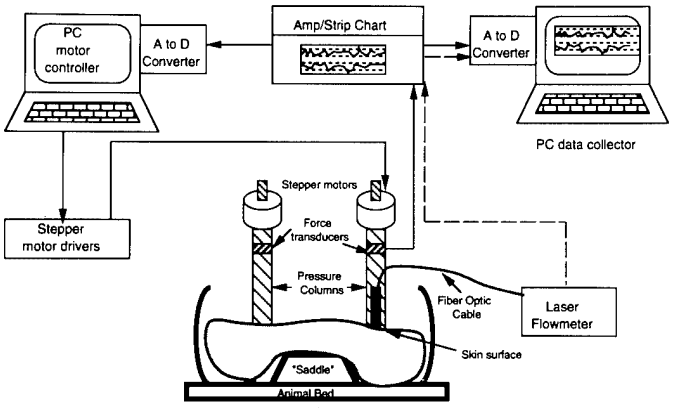

Figure 1a.
Schematic diagram of pressure delivery system and laser Doppler flowmeter. For simplicity, signals from only one pressure column are diagrammed. Dashed lines represent signals from the laser Doppler flowmeter.

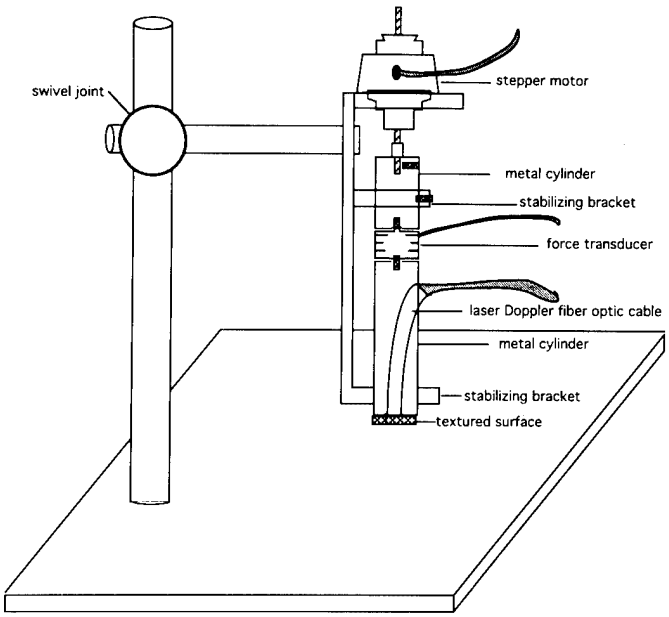

Figure 1b.
Detailed diagram of pressure column.

Apparatus

Tissue interface pressure is produced by using a closed-loop system that regulates the applied force (see **Figures 1a** and **1b**). In addition, pressure-induced reduction of cutaneous blood perfusion is monitored by using a fiber optic laser Doppler flowmeter (**Figure 1c.**) Pressure is applied to the skin by a metal column consisting of two stainless steel cylinders and a force transducer. The central portion of the column is fitted with a force transducer to measure the applied force. The force signal is also used by the computer to precisely maintain and deliver constant cutaneous pressure within a range of ±5 g of force (±2.9 mm Hg). At the tip of the second stainless steel cylinder is a shape indenter which is labeled "textured surface" in **Figure 1b.** This is the material that makes contact between the pressure column and the skin surface. Each of these components is discussed in detail below.

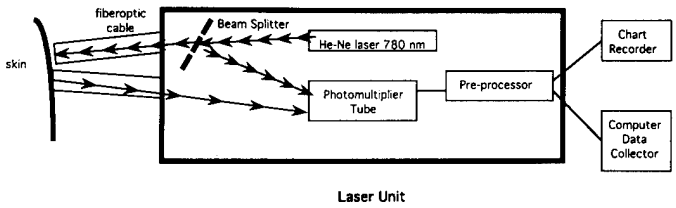

Figure 1c.
Detailed schematic diagram of laser Doppler flowmeter. The laser produces light with a wavelength of 780 nm. After penetrating the skin, the light beam is reflected back into the fiber optic cable. The beam splitter then separates the incident beam from the reflected signal beam. The preprocessing module subtracts the incident frequency from the reflected frequency to yield a value for the laser Doppler shift which is recorded by the strip chart recorder and saved on disk by the data collection computer.

Textured Surface

The textured surface or shape indenter is made from Safety-Walk® Tread (3 M Company, Cat.#7739, St. Paul, MN). This medium duty friction tape contains a pressure sensitive adhesive on one side (attached permanently to the pressure column) and on the other side contains a rubber-like, resilient material with a stippled surface that comes into contact with the skin.

Force Transducer

A 5-lb capacity single axis force transducer (load cell, model ALD-Mini-UTC-5, A. L. Design, Inc., Buffalo, NY) provides cutaneous surface pressure data to the computer. The force transducer is threaded on both ends, allowing it to be placed between two stainless steel cylinders, each 0.5 in (1.27 cm) in diameter. The force transducer and the two metal cylinders compose the pressure column (**Figure 1b**). The force transducer load cell requires 10 V excitation and delivers a rated output of 10 mV. The force transducer output is observed on a modified strip chart recorder (Servogor 124, Norma Goerz Instruments, Elk Grove Village, IL) for real-time hard copy output. The amplified signal from the chart recorder is sent to the analog to digital (A/D) converter (A/D card model number PCL-711, Laboratory Technology, Marlboro, MA) of a personal computer (stepper motor control computer) for feedback control of the tissue interface force. The strip chart recorder can also display flow measurements from the Perimed laser Doppler flowmeter (Periflux by Perimed, Piscataway, NJ) as described below. After A/D conversion (A/D card model DT2801, Data Translation, Marlboro, MA), the amplified force and flowmeter signals are recorded on a second personal computer with the aid of Perisoft software and saved to disk for future analysis.

The pressure column is calibrated using customized software (SMC8 version 1.3, Center for Biomedical Engineering, University of Kentucky, Lexington, KY). The pressure column is first inverted and zeroed with a weight. Known weights of up to 200 g are then placed on the flat surface of the metal column and the A/D counts are measured to establish the linear calibration for each transducer. The software then fits a linear relationship between counts and grams. When the column is turned right side up, the weight of the lower half of the column is zeroed, and the linear change in voltage per change in force as measured during the calibration procedure accurately indicates the force change. Calibration can also be achieved by having the column push against an appropriate scale.

Although this particular force transducer cell is designed to be used in a vertical position, we have demonstrated that it will function accurately when positioned in various angles so that we can best approximate perpendicular pressure to the tissue surface during the procedure. A 3-axis force cell is planned for future experiments in which shear force components are to be measured.

Stepper Motors

Delivery and maintenance of specified cutaneous pressure is accomplished by using highly reliable stepper motors to displace the metal column. The miniature stepper motors (model K92100, Phillips Technologies, Airpax Mechatronics Group, Charlotte, NC) were factory modified to act as bi-directional linear actuators. The motor shaft is a screw and the rotor is threaded on a low friction ball bearing in the motor casing. This makes it possible for the screw shaft to be moved in a bi-directional linear fashion as opposed to the more common clockwise and counterclockwise rotation without linear movement. A stepper motor driver chip (model SA1042V, Motorola, Phoenix, AZ) directly controls the firing sequence of each motor coil. Each 12-V direct current bipolar driver chip controls one stepper motor and receives two input signals via the printer port of the computer. The most significant bit of the computer signal controls the direction of the motor, and the least significant bit provides the step pulse. The motor driver chip accepts three input signals, contains a state controller, and provides two output signals

(22). The three driver input signals are "full/half step," "direction," and "clock." The "full/half step" option is set at half step to give the motor a resolution of 0.0005 in (0.00127 cm) of linear movement per step. The chip achieves the half step option by activating at least two of the four motor coils per step. One coil "pushes" and another coil "pulls" to prevent a full rotation of the threaded nut. The half step allows better resolution but sacrifices speed and draws more current than the full step option. The "direction" input level is set by an open collector buffer chip (model 7407, Texas Instruments, Newark Electronics, Chicago, IL) and serves the function of keeping the direction in a definite logic state while permitting transistor-to-transistor logic (TTL) compatibility with the computer (22). The computer signal for a step is connected to the clock input pin. Stepping is achieved on the rising edge of the clock pulse. The state controller converts the digital input signal from the computer into a linear motor output step. This output is provided in the two states of full or half step. The activity and control of the moment-to-moment activity of the motor and the changes in the status of the motor (moving up, down, or not changing) are monitored in a feedback loop using an A/D board and then viewed on the monitor by using customized computer software.

Interactive Software

The cutaneous pressure is maintained by a customized interactive software system (Center for Biomedical Engineering, Lexington, KY) that utilizes the signal from the force transducer to control the stepper motors. The interactive software provides the user with the choices of manual motor control or automatic tracking. In the automatic tracking mode, the user initially enters the desired target cutaneous pressure and the system automatically maintains the pressure within the user defined range. When pressure falls outside the established target range, the computer will signal the stepper motors to move the number of steps required to return to the target range. The system automatically adjusts to minor changes in rat body movement. The number of steps estimated by the computer is a function of the difference between the current pressure and the target pressure. The computer system is capable of responding to a change in pressure as small as 0.25 g (0.15 mm Hg). The cutaneous pressure range is set so that pressure variations secondary to body movements from the cyclical pattern of normal respiration do not trigger motor stepping adjustments.

Determination of Pressure-Induced Blood Perfusion Changes

Alterations in cutaneous blood perfusion are monitored using the Periflux Laser Doppler Flowmeter system. With this system, a fiber optic probe emits laser generated monochromatic light (780 nm) that is scattered and absorbed by the tissue being studied. This fiber optic probe is located in the center of the distal portion of the metal pressure column to monitor changes in relative cutaneous blood flow before, during, and after application of cutaneous pressure. Light reflected off blood cells moving through the scattered laser light will experience a change in wavelength known as the Doppler shift and will travel back to the "master unit" through a second fiber optic cable. The velocity and number of moving blood cells in the skin is related to the proportion of Doppler-shifted to non-Doppler-shifted light. The Doppler shift recorded by the master unit consists of the frequency difference between Doppler- and non-Doppler-shifted light.

The laser Doppler flowmeter is used to measure mean velocity of blood cells in a sample site of tissue, the number of blood cells moving in the sample site and blood cell perfusion (flux) which is the product of mean velocity and the number of blood cells. The sample depth is approximately 1 mm below the surface of the skin. Because of the sensitivity of the laser light to penetration, absorption, and reflection to tissue and blood parameters, the values obtained from the flowmeter are used for relative comparisons from controls in the same animal and not as absolute values.

The basic unit of measurement by the flowmeter is the perfusion unit (PU) or flux. One PU is equal to an analog output of 10 mV. The PU or flux does not represent a standardized value, such as the absolute number of cells moving through a given volume of tissue over a specific time. Rather, the PU is an arbitrary value relative to a specific application and depends on such factors as the wavelength of laser light, type of probe, temperature of subject, systemic blood pressure, posture, and biological zero perfusion value (23). Because of Brownian motion, a true zero reading for tissue perfusion measurements cannot be attained; however, by using a latex control suspension it is possible to compensate for much of the variation caused by this phenomenon. The corrected value is referred to as the biological zero. In the present system, the combined movement and concentration of blood cells through the cutaneous microvasculature (flux, PU) is determined for the central region under the pressure column (**Figure 2**).

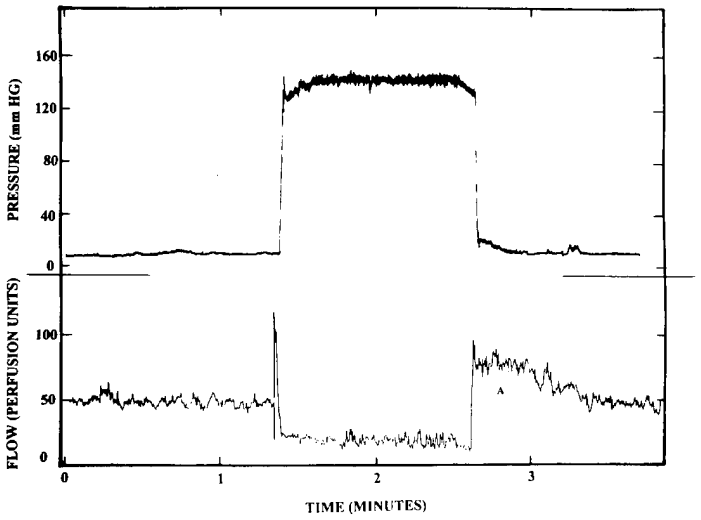

Figure 2.
The relationship between increasing pressure and decreasing blood flow. The upper chart shows pressure versus time while the lower chart shows blood flow versus time. Pressure increases from 5 to 250 g while blood flow as measured by the laser Doppler flowmeter decreases from its initial value to biological zero. A typical hyperemic response (A) occurs when pressure is released.

The mode of delivery of the pressure is depicted in **Figure 3**; a line drawing of the hip of the rat that shows where cutaneous pressure is applied. Animals are routinely placed in the base of the device and the angle orientation of the pressure column is directed at the same location of the hip (greater trochanter) each time, as previously described. Pressure ulcers form in the trochanter region of the rat above the bony prominence, which is the area covered by the pressure column during the procedure.

Reliability of Measurements

On a daily basis, anesthetized subjects are placed in the device and pressure is applied, an initial zero baseline reading of the force cell having already been determined. The force cell output is recorded on the strip chart recorder. The feedback regulation of the stepper motors is accomplished by using the motor controlled PC system. In addition, a strip chart recorder and computer-based storage system are used to document the maintenance of cutaneous pressure throughout the 6-hr session (**Figure 1a**).

Prior to the evaluation of the impact of pressure on cutaneous ulcer development, a series of control studies was conducted to characterize the reliability and accuracy of the feedback system. **Figure 4** demonstrates the maintenance of reliable pressure by using an inert substance

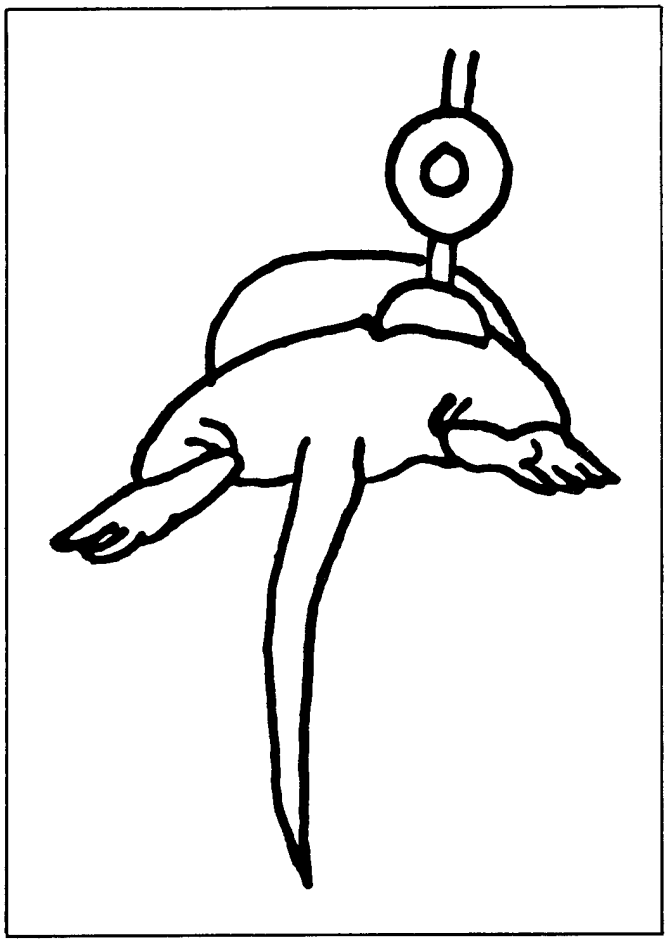

Figure 3.
Line drawing showing the contact between the pressure column and the trochanter region of the fuzzy rat.

(foam rubber) as a model to validate the feedback control of the stepper motors. **Figure 5** demonstrates that reliable pressure can be maintained by the computer-controlled system over an extended period of time (2 hrs). Similar plots were achieved (not shown) for periods in excess of the typical 6-hr pressure session. To control for various physical compliance/elasticity aspects of normal skin and underlying tissue, the system was tested by using a recently euthanized rat (data not shown). Despite slight, momentary shifts, steady pressure (145 mm Hg ±2.9 mm Hg) can be maintained by the computer system for extended periods of time. This result demonstrates that the computer-controlled system can reliably alter the pressure column via the stepper motor to maintain the target pressure within preset limits.

After the initial characterization using a non-living model system (**Figure 4**), an anesthetized rat was placed

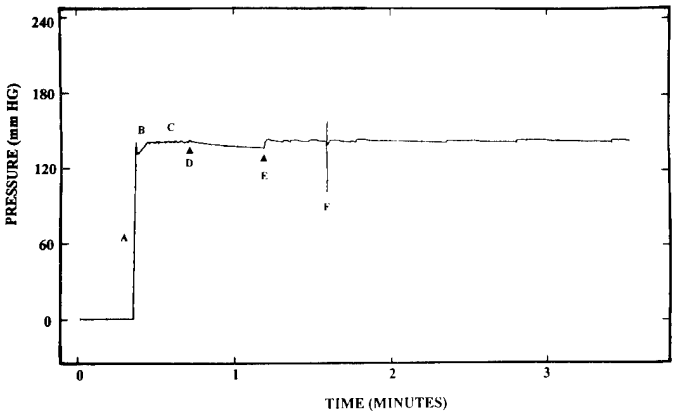

Figure 4.
Typical result of force application to an inert surface. The test protocol represents the application of 250 g of force to foam rubber. After the desired force is typed in, the system produced a rapid rise in force (A) followed by a more gradual displacement (B) until the desired force is reached (C). Small corrections made by the controller can be seen because of the creep characteristics of the foam material. For example, when the controller is turned off (D) the force decreases because of material creep. The desired force is returned when the controller is turned on (E). The system's ability to recover from a displacement disturbance is also shown (F).

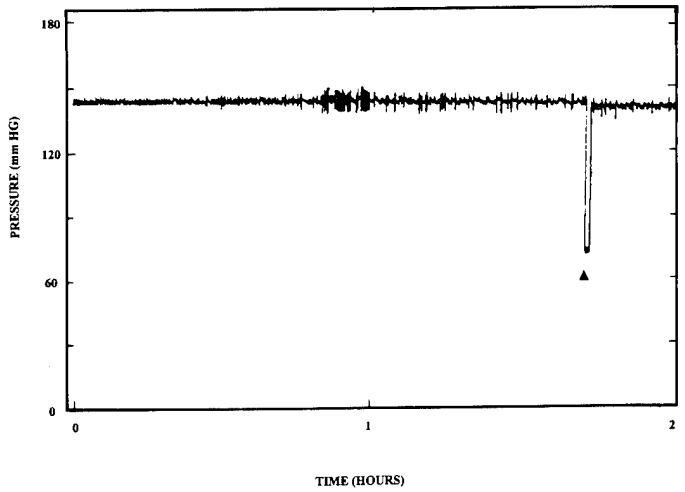

Figure 5.
Typical plot of force application of 250 g for 2-hr duration. Demonstration of the ability of the system to compensate for respiration and other factors. The large spikes represent actual motor adjustments. The smaller spikes represent respiratory changes which do not trigger a motor response. The arrow points to our artificial mark indicating the addition of anesthetic.

into the restraining device and the characterization of pressure induced perfusion changes in the cutaneous compartment below the pressure column was determined. **Figure 2** presents a 4 min expanded scale on the x axis, showing the moment-to-moment movement in the force tracing when pressure brought tissue perfusion to "biological zero " (i.e., no flow as demonstrated by the laser Doppler flowmeter) and is a typical perfusion response to pressure. The initial baseline perfusion when no pressure is applied demonstrates a high rate of cutaneous blood flow. As the pressure on the skin is increased by the movement of the column, as controlled by the stepper motor, there is a progressive reduction in perfusion blood flow to an apparent level of zero. This level of ischemia is maintained with a pressure of 145 mm Hg until the pressure is released, resulting in a hyperemic response (see "A" in **Figure 2**).

Minor changes in the pressure delivery occur as a result of movements associated with the animal's respiratory cycle (see **Figure 5**) and are ignored by presetting the range of acceptable values. When force values become too large, the feedback mechanism of the system returns the pressure to within the preset limits as described in the methods section. It is important to note that this particular design of a feedback control stepper motor system is able to reliably maintain cutaneous pressure and minimize blood flow under the pressure column for the entire 6-hr session.

Physiological Measurements

After two fuzzy rats were anesthetized, the left carotid arteries were cannulated in order to allow for the intermittent measurement of mean arterial blood pressure, and heart rate (24). The animals were not subjected to pressure sessions.

RESULTS

After repeated exposure to daily pressure for 6-hr periods (5 consecutive daily sessions), rats develop macroscopic lesions associated with cutaneous ulceration. The pressure ulcers form directly under the area covered by the pressure column. As shown in **Figure 6,** there is a marked change in the appearance of postpressure skin as compared to normal skin in the rat. Stage 2 pressure ulcers show microscopic evidence of infarction with infiltration of white cells along the margin of the necrotic tissue (**Figure 7a**) and necrosis of tissue with the appearance of underlying emerging liquefactive necrosis. There

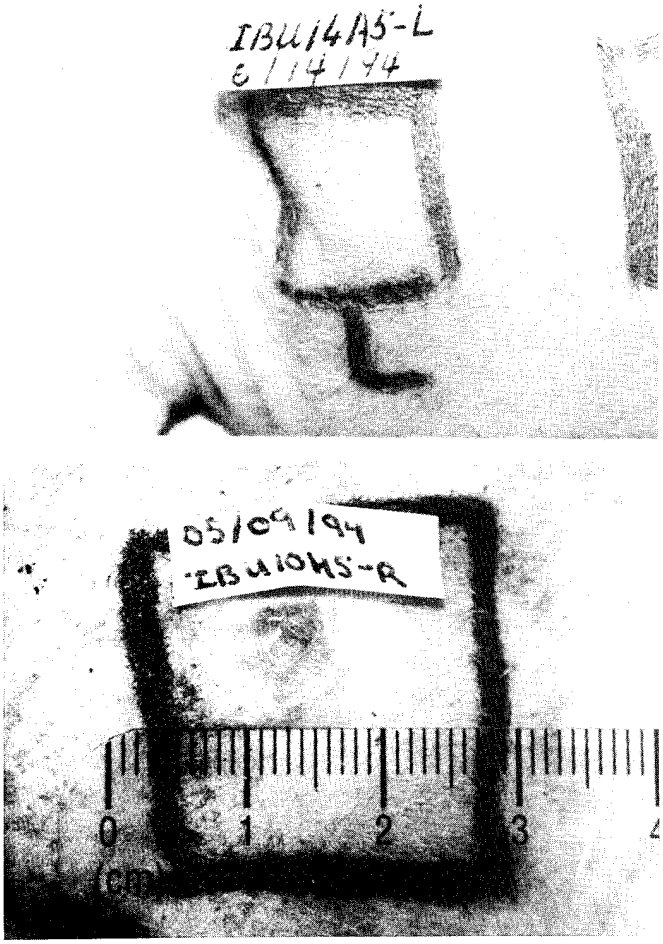

Figure 6.
Light macrographs of normal (top) and pressure treated skin (bottom) of the fuzzy rat.

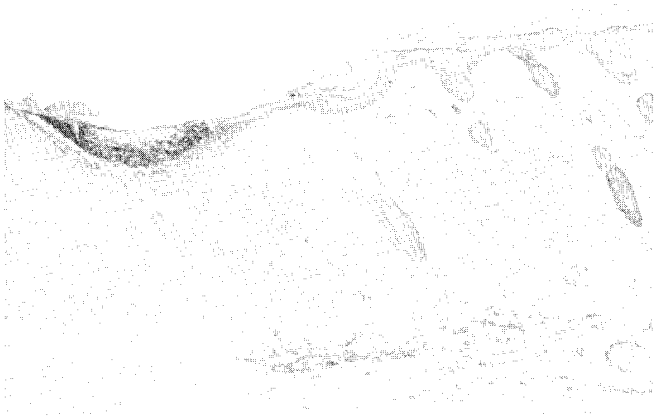

Figure 7a.
Light micrographs of damage caused by the application of pressure to the skin of the fuzzy rat: an infarct to the skin.

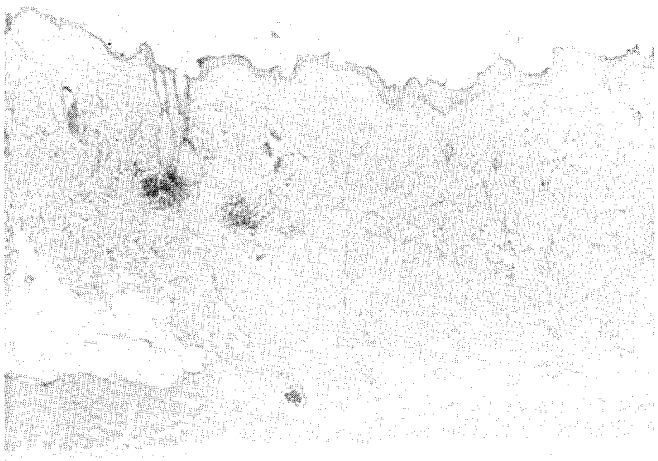

Figure 7b.
Light micrographs of damage caused by the application of pressure to the skin of the fuzzy rat: panniculus carnosus damage.

is evidence of edema and margination of white cells at the level of arteriolar supply of the muscularis underlying the skin. A common early and/or mild lesion was necrosis of the panniculus carnosus muscle and the superficial adipose tissue (**Figure 7b**). Thrombi were occasionally noted (**Figure 7c**). At this stage in the development of the lesion, it appears that the primary site of lesion development is in the subcutaneous muscle layer. To date, we have successfully used this system on more than 300 rats to evaluate hypotheses on the origin and treatment of pressure ulcers. Many of these rats were used in refining the histopathology of experimental ulcers, or in flow studies. Many were used in studies verifying the model and in exploring the models' utility in testing various drug interventions.

In order to further validate the model system, a single experiment was designed to elucidate the relationship between blood flow and pressure in the fuzzy rat. This experiment generated the flow versus pressure curve shown in **Figure 8.** Tissue perfusion blood flow decreased nonlinearly with increasing skin pressure and approached zero flow at 35–40 mm Hg, which is approximately equal to human capillary pressure. Tissue perfusion was completely cut off at 80 mm of Hg. A best fit curve analysis of

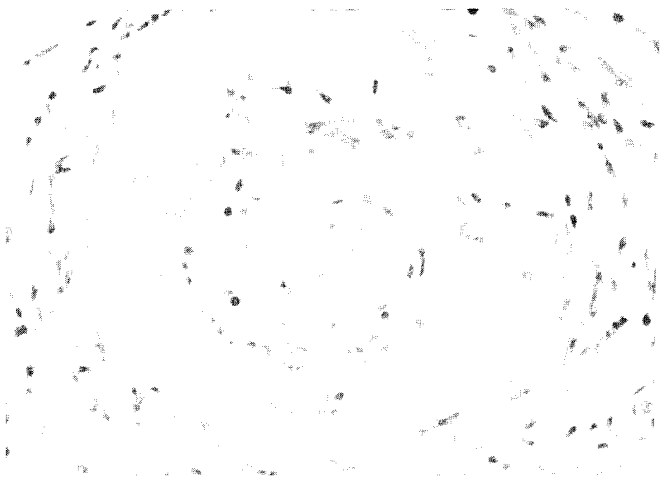

Figure 7c.
Light micrographs of damage caused by the application of pressure to the skin of the fuzzy rat: occlusion of an arteriole.

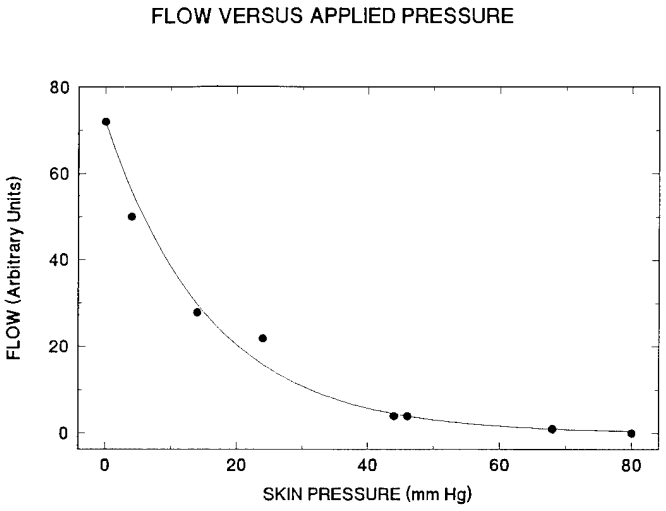

Figure 8.
The relationship between increasing pressure and decreasing tissue blood flow as measured by the laser Doppler flowmeter. Pressure ranges 0–80 mm Hg (0 to 137 g force). The variable application of force was over a 5-min time period.

the data points by Axum (TriMetrix, Inc., Seattle, WA) and SigmaStat (Jandel Scientific, San Rafael, CA) software results in the exponential curve shown.

Table 1 shows the physiological measurements of mean arterial blood pressure (mm Hg) and average heart rate (beats per minute) for two fuzzy rats over a 24-hr time span. The time values begin at the point in time when a single i.p. injection of anesthesia was given. During the period in which the rats remained unconscious, mean blood pressure values ranged from 112–124 mm Hg, while average heat rate varied from 440–492 beats per minute. During the recovery period, mean blood pressure increased slightly to a range of 132–140 mm Hg, and the average heart rate varied from 424–508 beats per minute. At 24-hrs postinjection, the mean blood pressure reading for both animals was 120 mm of Hg, and the average heart rate had settled into a range of 360–376 beats per minute.

DISCUSSION

This project has developed a reliable animal model and surface pressure delivery system for the study of pressure ulcers in the laboratory environment. The animal model described in this paper differs from models in previous studies, in that it uses a computer-controlled system capable of making perfusion measurements during the application of precisely controlled tissue pressure, thus allowing experimental control and monitoring of some of the crucial variables central to the study of pressure ulcer development. These critical variables include pressure, temporal blood flow, interval histopathology, and biochemistry. Furthermore, the model has utility for the study of pharmacologic intervention.

Our use of 250 g force or 145 mm of Hg is almost 2 times higher than the amount of force required to completely cut off tissue perfusion (80 mm Hg) as measured by laser Doppler flowmeter. This application of 250 g force for 5 consecutive daily sessions of 6-hrs duration each, consistently results in the formation of experimental pressure sores (greater than a 90 percent incidence).

The selection of 145 mm Hg and the duration time of 6 hrs per session was based on a review of the pressure ulcer literature and the results of the pressure versus flow experiment (**Figure 8**). The model is extremely flexible, in that pressure and/or duration may be varied, as demonstrated by **Figure 8.** We are currently investigating various pressure and time combinations in order to further elucidate the model and to clarify the relationship between pressure and time in the etiology of experimental pressure ulcers.

The mean arterial blood pressure readings in fuzzy rats of 120 mm Hg at 24-hrs postanesthesia injection are within the normal range of 115–121 mm Hg reported for the conscious rat (24). After 24 hrs, the animals have had time to recover from both anesthesia and the cannulization

Table 1.
Mean blood pressure and heart rate measurements in the fuzzy rat.

Time (HR)[a]	Rat 1[b]		Rat 2[c]	
	Mean Blood Pressure[d] (mm Hg)	Heart Rate[e] (BPM)	Mean Blood Pressure[d] (mm Hg)	Heart Rate[e] (BPM)
0.5	112	484	120	460
1	120	448	128	464
2	120	440	124	492
3	132	424	140	508
4	132	448	136	496
5	124	384	116	376
6	128	392	116	328
7	124	416	124	352
24	120	376	120	360

a=Hours after the i.p. injection of the anesthesia mixture; b=Male fuzzy rat weighing 360 grams; c=Female fuzzy rat weighing 244 grams; d=Mean arterial blood pressure is defined as the value of 2 times the diastolic pressure plus the systolic pressure divided by 3; e=BPM equals beats per minute.

procedure. As one might expect, the mean blood pressure and heart rate values show less variation under anesthesia. The mean blood pressure values during anesthesia and the conscious state were similar. The force used to generate experimental pressure ulcers in this study is only slightly greater than the mean arterial pressure of the anesthetized or conscious fuzzy rat. Externally applied pressure as a pressure ulcer causative factor may be significant largely in its relationship to the internal blood pressure of the animal. The relationship between externally applied pressure and internal blood pressure as revealed in this study, and its subsequent impact on the generation of pressure ulcers is well within theoretical expectations.

Advantages of the Fuzzy Rat

The use of the fuzzy rat provides several advantages. This rat is hypotrichotic (21) and therefore does not require depilation or prior treatment of the skin which might otherwise result in artifacts during the formation of pressure ulcers. The rat is inexpensive in both initial cost and maintenance and therefore highly efficient from a cost standpoint and well-suited for large experimental trials. In addition, more is known about the pharmacological effects, the absorption, distribution, and metabolism of drugs with the rat than with any other species, including the human.

The Long-Term Utility of This Model

The model presented in this paper permits the investigator to continuously monitor the exact pressure delivered to the skin. The force transducer provides moment-to-moment data on the pressure delivered and the laser Doppler flowmeter provides concurrent information on tissue perfusion. In this regard, it is possible to determine pressure-flow relationships in the development of pressure lesions and to test hypotheses relating to this interaction. With respect to evaluating potential therapeutic approaches (e.g., drugs), the ability to monitor relative changes in blood flow to tissues during pressure delivery makes it possible to control for drug-induced changes in blood flow to tissue as a potential mechanism of protection. In addition to tissue surface pressure and tissue perfusion, interval histopathology and biochemistry are among the critical variables central to the study of pressure ulcer development.

By using our model, the pathophysiologic continuum observed in the development of a pressure ulcer can be monitored and evaluated. Components of this continuum and their influence can be studied separately and in concert. Additionally, the development of this model will allow us to test pharmacologic compounds and mechanisms that have a potential use in the prevention and treatment of pressure ulcers, including anti-inflammatory drugs, angiogenic agents, growth factors, fibrogenic drugs, antioxidant compounds, free radical scavengers, and specialized drug delivery systems.

Although this model allows for manipulation and control of the critical variables in the laboratory environment regarding the pathophysiology of pressure ulcers, it lacks the natural history and the elements found in the human environment. However, we will be able to test cer-

tain hypotheses in order to gain a more fundamental approach to study the complex process involved in the development of pressure ulcers. The model mimics the human environment in several key respects; thus, the results of our research may be generalized to the human condition. The development of histopathological changes (lesions) following pressure application has been the focus of several studies utilizing different models of pressure sore development. Husain (13) reported microscopic changes in rat muscle subjected to pressure for periods of 1–10 hrs. Keane (17) reported that muscle is more susceptible to pressure damage than skin in the immobilized patient. Using the model described in this paper, we have confirmed these results (25) and presented data suggesting that these lesions may be initiated by postischemic free radical production (26) analogous to what has been reported in brain and cardiac tissues following ischemia-reperfusion injury (27). As indicated in the histopathology figures, some of this damage may be initiated by neutrophil-mediated activation and infiltration of the tissue. In this process, activated neutrophils generate reactive oxygen species and injure cells. Although Koziak's study (5) gave evidence that lesions first developed in the dermis and then eroded downward, our results are more in agreement with the findings of Husain (13) and Keane (17), that lesions originate in the muscle, and that muscle may be more sensitive to ischemia than skin. Our findings show that muscle is very sensitive to pressure injury and first to show injury; however, the lesion that becomes an ulcer may develop at all depths simultaneously. We also have a second, invasive, laser Doppler flow probe to measure muscle blood perfusion, before, during, and after the application of tissue pressure to further evaluate this point. Although these observations suggest a role of free radical and white cell mediated damage, mechanistic studies remain to be conducted.

CONCLUSIONS

This novel multisystem approach in the scientific study of pressure ulcers will serve as a basis for a better understanding of the molecular biology and pathophysiology leading to the development of pressure ulcers in a controlled environment. Although other animal/human models with various types of pressure surface delivery devices (5,12,16,17,28) are available, some with computer and/or closed loop feedback control (14,20), we believe that the model system presented here, is the smallest, most efficient and cost effective system for large scale studies of pressure ulcer development in a laboratory controlled environment.

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REFERENCES

1. Allman RM. Pressure sores among the elderly. N Engl J Med 1989:320:850–3.
2. Bergstrom N. A research agenda for pressure ulcer prevention. Decubitus 1992:5:22–30.
3. Landis EM. Micro-injection studies of capillary blood pressure in human skin. Heart 1930:15:209–28.
4. Trumble HC. The skin tolerance for pressure and pressure sores. Med J Australia 1930:2:724–6.
5. Kosiak M. Etiology and pathology of ischemic ulcers. Arch Phys Med Rehabil 1959:40:62–9.
6. Lindan O, Greenway RM, Piazza JM. Pressure distribution on the surface of the human body: evaluation in lying and sitting positions using a "bed of springs and nails." Arch Phys Med Rehabil 1965:46:378–85.
7. Kosiak M, Kubicek WG, Olson M, Danz JN, Kottke FJ. Evaluation of pressure as a factor in the production of ischial ulcers. Arch Phys Med Rehabil 1958:39:623–9.
8. Barnett RI, Ablarde JA. Skin vascular reaction to standard patient positioning on a hospital mattress. Adv Wound Care 1994:7:58–65.
9. Lee EB, Anhalt GJ, Voorhees JJ, Diaz LA. Pathogenesis of scleroderma (Current Concepts, Review). Int J Dermatol 1984:23:85–9.
10. Ferguson-Pell MW. Seat cushion selection. J Rehabil Res Dev Clin Suppl 1990:2:49–73.
11. Blaylock B. Pressure ulcers: a review. Dermatol Nurs 1990:2:278–82.
12. Groth KE. Clinical observations and experimental studies of the pathogenesis of decubitus ulcers. Acta Chir Scand Suppl 1942:76:1–209.
13. Husain T. An experimental study of some pressure effects on tissues, with reference to the bed-sore problem. J Pathol Bacteriol 1953:66:347–58.
14. Daniel RK, Priest DL, Wheatley DC. Etiologic factors in pressure sores: an experimental model. Arch Phys Med Rehabil 1981:62:492–8.

15. Harman JW. The significance of local vascular phenomena in the production of ischemic necrosis in skeletal muscle. Am J Pathol 1948:24:625–41.
16. Nola GT, Vistnes LM. Differential response of skin and muscle in the experimental production of pressure sores. Plastic Reconstr Surg 1980:66:728–35.
17. Keane FX. The function of the rump in relation to sitting and the Keane reciprocating wheelchair seat. Paraplegia 1978:16:390–402.
18. Dinsdale SM. Decubitus ulcers in swine: light and electron microscopic study of pathogenesis. Arch Phys Med Rehabil 1973:54:51–6.
19. Hagisawa S, Ferguson-Pell M, Cardi M, Miller SD. Assessment of skin blood content and oxygenation in spinal cord injured subjects during reactive hyperemia. J Rehabil Res Dev 1994:31(1):1–14.
20. Ferguson-Pell M, Hagisawa S, Masiello RD. A skin indentation system using a pneumatic bellows. J Rehabil Res Dev 1994:31(1):15–9.
21. Palm J, Ferguson FG. Fuzzy, a hypotrichotic mutant in linkage group I of the Norway rat. J Hered 1976:67:284–8.
22. Stepper motor specifications, series 92100 Digital Linear Actuator Technical Sheets, Charlotte, NC: Phillips Technologies, Airpax Division, 1992.
23. Periflux 4001 master user's manual, Perimed AB, Jarfalla, Sweden, 1992.
24. Dowell RT, Kauer CD. Uteroplacental blood flow at rest and during exercise in late-gestation conscious rats. J Appl Physiol 1993:74:2079–85.
25. Salcido R, Donofrio JC, Fisher SB et al. Histopathology of pressure ulcers as a result of sequential computer-controlled pressure sessions in a fuzzy rat model. Adv Wound Care 1994:7:23–40.
26. Salcido R, Carney JM, Fisher SB, LeGrand EK, Dickey KM, Liu L. A reliable animal model of pressure sore development: the role of free radicals (Abstract). J Am Paraplegia Soc 1993:16:61.
27. Cao W, Carney JM, Duchon A, Floyd RA, Chevion M. Oxygen free radical involvement in ischemia and reperfusion injury to brain. Neurosci Lett 1988:88:233–8.
28. Swaim SF, Bradley DM, Vaughn DM, Powers RD, Hoffman CE. The greyhound dog as a model for studying pressure ulcers. Decubitus 1993:6:32–40.

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Experiments in dysarthric speech recognition using artificial neural networks

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Abstract—In this study, we investigated the use of artificial neural networks (ANNs) to recognize dysarthric speech. Two multilayer neural networks were developed, trained, and tested using isolated words spoken by a dysarthric speaker. One network had the fast Fourier transform (FFT) coefficients as inputs, while the other network had the formant frequencies as inputs. The effect of additional features in the input vector on the recognition rate was also observed. The recognition rate was evaluated against the intelligibility rating obtained by five human listeners and also against the recognition rate of the Introvoice commercial speech-recognition system. Preliminary results demonstrated the ability of the developed networks to successfully recognize dysarthric speech despite its large variability. These networks clearly outperformed both the human listeners and the Introvoice commercial system.

Key words: *artificial neural networks, cerebral palsy, dysarthric speech, speech recognition.*

INTRODUCTION

While users need not have "normal or perfect" speech to exploit available speech recognition systems, the input speech must be consistent. This usually seems impossible for individuals with cerebral palsy because of their lack of control of articulatory movements during speech production. The inconsistency in dysarthric speech precludes its recognition by currently available commercial systems (1). Miller et al. (2) analyzed the speech variations in utterances produced by individuals with cerebral palsy in a speech recognition study using the Dragon VoiceScripe system. They reported that the accuracy of the system was extremely dependent on the repeatability of voice commands in the tone and spectral content. In another study, using the Interstate Voice Products' speech recognition system with a dysarthric speaker afflicted with cerebral palsy, Lee et al. (3) reported that, even with retraining, the subject only reached an overall accuracy of 70 percent. Carlson and Bernstein (4) reported a wide range of percentage of recognition from speaker to speaker in a study involving 50 subjects with articulation disabilities, mainly with hearing impairment and cerebral palsy. The system was more successful for the subjects with hearing impairment than for the subjects with cerebral palsy.

Goodenough and Rosen (5) reported that speech recognition performance rapidly deteriorated for vocabulary sizes greater than 30 words, even for persons with mild to moderate dysarthria. It is likely that individuals whose dysarthria is severe enough to derive benefits from using augmentative communication devices may not be expected to gain much from these commercially available speech recognition systems. There are also various disadvantages to these commercial systems; such as, the requirement for repeatable word patterns between training and operation, the inability to cope with ambient

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noise, and inadequate interfaces with rehabilitation devices (2).

Recent research has focused on the assessment of dysarthric speech and the utility of computer-based speech recognition systems. Sy and Horowitz (6) described a causal model which addressed important issues; such as, using normal speech as a control in evaluating dysarthric speech, categorizing speech errors in terms of its features, and determining the relationship between intelligibility rating and speech recognition performance. Coleman and Meyers (7) examined computer recognition of dysarthric speech through the use of a structured model and concluded that the low overall recognition rate of dysarthric speakers remains a serious problem. These researchers suggested that this problem could be approached in two ways: changing the speech input signal and changing the recognition system.

Changing the speech input signal possibly might be achieved through speech training and therapy, but this is unlikely. Changing the recognition system, on the other hand, involves the development of more robust techniques to handle the variability and inconsistency in dysarthric speech. Possibilities include the development of algorithms that filter out sounds beyond a certain length and at certain frequencies. Another possible technique is the use of artificial intelligence through which the algorithm can learn and compensate for the types of inconsistencies produced by the speaker (7). A third technique is based on the modeling of useful parts of cerebral palsy speech; such as, vowel-like strings punctuated by inappropriate (compared to normal speech patterns), sounds and silences. Using this technique, known as hidden Markov model (HMM), an overall recognition rate of 90 percent was reported for vowel sounds (8). Similar results were also obtained by Boonzaier and Limon (9).

The use of HMMs, despite their relative success, has many limitations. These include poor low-level acoustic modeling and poor high-level semantic modeling. Poor low-level acoustic modeling leads to confusions between acoustically similar words, while poor high-level semantic modeling restricts applications to simple situations with limited vocabulary. Also, HMMs do not model coarticulation directly and cannot model the topological structures of words and subwords (10). These limitations are more pronounced for cerebral palsy speech because of its high degree of variability. In addition, HMM theory does not specify the structure of implementation hardware which is important for interfacing with rehabilitation devices.

The use of artificial neural networks in speech recognition provides the potential to overcome these limitations. Neural networks can perform better than existing algorithms because they adapt their internal parameters over time to maximize performance and self-organize to capture new features as they are observed (11). During training, the neural networks successively update information learned from past experiences, giving them the ability to handle variations and inconsistencies in the speech signal and to process incomplete or missing data (12). Lerner and Deller (13) pointed out that neural network structures may hold promise for recognition of cerebral palsy speech. They introduced a neural network approach to learning invariant spectral features in cerebral-palsied speech.

This approach was adopted in the present study. First, an attempt was made to identify the range of variability in dysarthric speech (a complete account of dysarthric speech can be found in references 14 and 15). Since cerebral palsy is the most common cause of dysarthria, a subject with cerebral palsy was selected for this purpose. It should be emphasized that many of the features of cerebral palsy speech (e.g., variability) are also features of dysarthria resulting from traumatic brain injury, stroke, or multiple sclerosis. Our subject's dysarthric errors, therefore, may also be found in the dysarthric speech of others with neurogenic communicative disorders. Second, we investigated the development of a high-performance recognition system for dysarthric speech using the technology of artificial neural networks.

METHODS AND MATERIALS

Subject

JK, aged 33, has cerebral palsy. His physical conditions include quadriplegia, spasticity, and athetosis. He has a bachelor's degree in sociology. He has used a direct selection device for 2 years (Light Talker made by Prentke-Romich Co., Wooster, OH). This device has a light-pen selector attached to a head band and a speech synthesizer to produce the selected message. Using this device, he can communicate an average of five words/minute. He is assisted in writing by a scribe who is familiar with his speech. As a child, he received speech therapy for about 10 years. His intelligibility score is 10–20 percent for average listeners and about 60 percent for people who are familiar with him. When asked to produce 25 multisyllable com-

mands such as "RETURN," "CLEAR," "RIGHT," and "DIRECTORY," a recognition rate of 20 percent was obtained using the IntroVoice speech recognition system. He uses a joystick to control his electric wheelchair. On several occasions, we met with JK to discuss the proposed system. He thought that it would be better for him to have a speech-recognition system. He said it would be faster than his device. "I like to talk," he added.

Speech Materials

The vocabulary for this study was selected using the following criteria: 1) use of monosyllabic words to initially simplify analysis; 2) inclusion of all vowel phonemes; 3) number of required words minimized (for the subject's convenience and to limit the amount of data to be analyzed); 4) words have a real-world application for the client such as augmentative communication or environmental/wheelchair controls; and 5) words are easily recognizable by the subject and have only one normal pronunciation. From a list of 50 words, the client made a short list of 20 words with which he was comfortable. Each word in the vocabulary was repeated 22 times. **Table 1** gives the list of the words used in this study.

Speech Processing

The recorded speech was amplified using Realistic SA-150 equipment. The output of the amplifier was then fed to an adjustable analog filter, a Krohn-Hite Model 3850. This acted as the anti-aliasing filter and was set to low-pass at 5 kHz. The output of the filter was next passed to the Data Translation DT2821 analog-to-digital converter. The sampling rate was set at 10 kHz.

A DOS batch file using ILS (16) software was used to effect segmentation of the words (i.e., finding the beginning and ending of a word). Each utterance was normalized to 45 frames at 256 points per frame. The segmented data were stored on an 80386, 40 MHz computer under appropriate names to distinguish the utterance and the repetition number, token, of that particular utterance.

Table 1.
List of the words used.

No.	Word	No.	Word
1	ONE	11	GO
2	OFF	12	START
3	HOW	13	FIVE
4	I	14	HAVE
5	WHY	15	WHAT
6	NO	16	HOME
7	PAIN	17	SIX
8	STOP	18	TURN
9	SAD	19	WHO
10	FOUR	20	ON

Feature Extraction

The fast Fourier transform (FFT) coefficients and the formant frequencies were extracted from all segmented data. The FFT were obtained by applying an eighth order (256 point) FFT to each frame of the segmented data. Only the real magnitudes of the FFT were used. Each frame provided 128 data points which were reduced to 16 points using the Turning-Point algorithm (17). The frequency, amplitude, and the bandwidth of the formants were extracted using linear predictive coding (LPC) analysis (18). The Interactive Laboratory System (ILS) was used to segment the speech signal and compute the linear predictive coefficients. These coefficients were next extracted using a program written in C language and then stored in separate files. The energy level of each frame was also provided by this program. The energy level was used to test the effect of additional features on the recognition rate.

Network Design

Two multilayer neural networks (19) were developed, trained, and tested using NeuralWare's Professional II/Plus package (20). One network had the FFT coefficients as inputs, while the other network had the formant frequencies as inputs. Both networks had hetero-associative, feed-forward, fully connected network configurations using a back-propagation learning algorithm and sigmoid transfer function. The main parameters of the network, number of layers, learning coefficients, and momentum were maintained the same so as to facilitate comparison of the recognition rate obtained by the two networks.

FFT Network

This network consisted of four layers: an input layer, an output layer, and two hidden layers. The input layer had 720 processing elements (PE). These PEs correspond to the 720 elements (16 elements/frame, 45 frames) in the input vector, representing each utterance. The first hidden layer had 270 PEs and the second had 90 PEs. The number of PEs in the output layer was determined by the num-

ber of words in the vocabulary (i.e., 20). **Figure 1** gives a schematic diagram of the neural network used in this study.

Formant Network

All the parameters in the FFT network were maintained except for the number of PEs, which was determined by the number of elements in the input vector. For the formant network, the input layer had 645 PEs, hidden layer 1 had 258 PEs, hidden layer 2 had 86 PEs, and the output layer had 20 PEs.

Network Training

The network was trained using the method of supervised learning. The data were presented to the network which produced an output. The difference between the actual output and the desired output was calculated and fed back to change the connections between the processing elements. From a total of 22 tokens per word in the vocabulary, a training set and a testing set were created. The training set consisted of 18 tokens/word, and the testing set consisted of 4 tokens/word. The separation of the tokens into training and testing sets was done randomly. The training set was further broken down to produce subsets with 6, 9, 12, 15, and 18 tokens.

Network Testing

The accuracy of the speech recognition network is defined by the *recognition rate*, which is the percentage ratio of recognized tokens to the total number of tokens used in testing the performance of the network. To find the optimum number of training iterations, the network was saved every 1,500 iterations for the first 20 check points, and from then on saved at every 10,000 iterations up to 100,000 iterations. The effect of increasing the number of training tokens on the recognition rate was also studied.

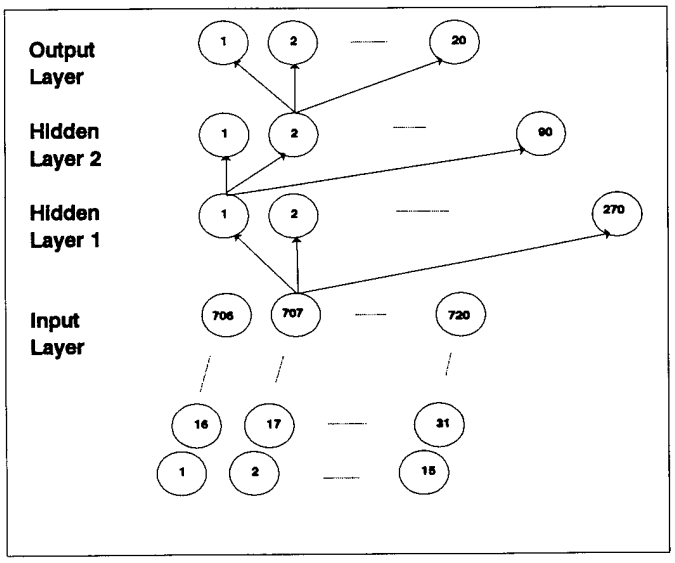

Figure 1.
Schematic of the neural network.

System Evaluation

The performance of our system was evaluated by comparing its recognition rate to the recognition rate obtained by the Introvoice speech recognition system. The recognition rates were also compared with the intelligibility ratings obtained on the subject's speech by five listeners with normal hearing. The intelligibility rating was obtained by using the Modified Rhyme Test (21,22) which is an ANSI standard (23) for testing intelligibility. This was a completion type, wherein the five listeners were provided with the stem of a word and then asked to fill in the first letter of the word they heard. Each error made in recognition of the initial consonant provided a clue to what kind of error the speaker made (i.e., a placement error, voicing error, and so forth). The order of the listening task and the replay of the speech recordings were randomized. Each listener was asked to select one word from a set of six rhyming words. The percentage of words correctly identified by each listener determined the intelligibility score. The average intelligibility score for the five listeners was then calculated for the speaker.

RESULTS

FFT Network

Figure 2 shows the variation of recognition rate with an increase in the number of iterations. This was for a network trained with 18 tokens. The 100 percent recognition rate was obtained when the network was tested using the training set. For the standard testing set (different sets were used for training and testing), the recognition rate improved as the number of iterations increased. A peak recognition rate of 76.25 percent was reached at 13,500 iterations, after which the rate dipped slightly to 75.25 percent and saturated thereafter.

Figure 3 shows the variation of the recognition rate as the number of training tokens increased. A gradual improvement was observed with the increase in the number

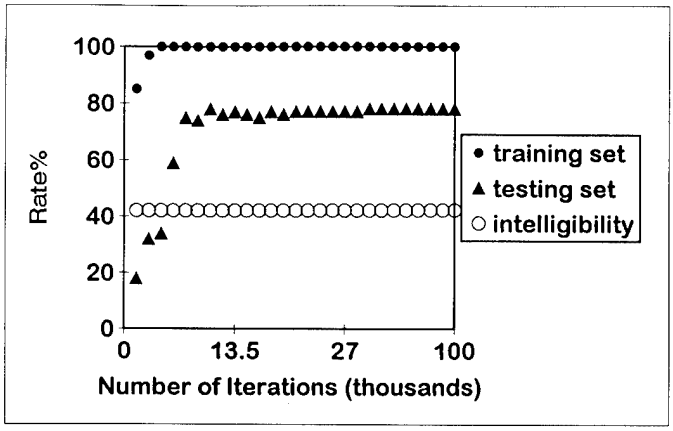

Figure 2.
Recognition results for FFT network.

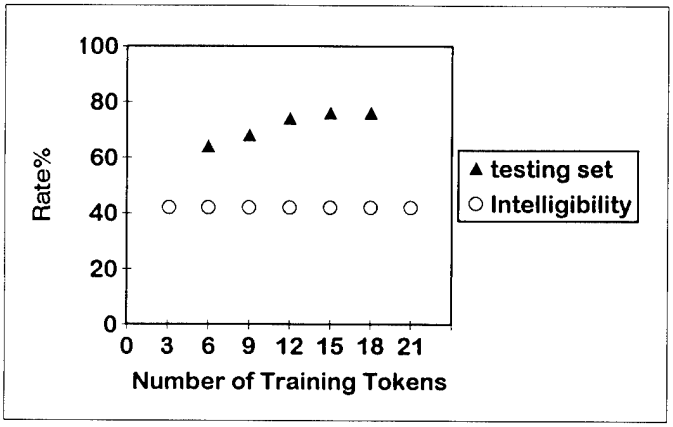

Figure 3.
Effect of training for FFT network.

of training tokens. A peak recognition rate of 76.25 percent was obtained for 15 training tokens and 18 training tokens. The activation levels of the two networks were studied to select the better network. In this experiment, the activation level was calculated as the difference between the highest activated node and the value of the second highest activated node in the output layer. This gave a measure of confidence with which a particular word was recognized. The network trained with 18 tokens had more recognized words falling in the higher confidence region than the network trained with 15 tokens. Hence, 18 was selected as the optimum number of training tokens in this experiment. The number of training tokens becomes more critical in dysarthric speech, considering the limited ability of the client to produce utterances without fatigue affecting his speech production.

The confusion matrix of the network trained with 18 training tokens is shown in **Table 2**. It shows that only the words 'go' and 'turn' (i.e., words 11 and 12) had low recognition rates. This demonstrated the inability of the subject to articulate these two words. The subject indicated that he was not comfortable producing these two words. This led us to further investigate his speech patterns before final development of the system.

The energy level of the speech signal was next added to the input vector to study the effects of additional features on recognition rate. The training set with 18 training tokens was used and a peak recognition rate of 78.25 percent was obtained. This represented an improvement of 2.25 percent over the peak recognition rate obtained before.

Formant Network

Figure 4 shows the recognition rate of the formant network trained with 18 tokens. This figure shows that a peak recognition rate of 42.5 percent was obtained. This indicated that the formant frequencies were not able to accurately track the variations in dysarthric speech as did the FFT network.

Figure 5 gives the peak recognition rate as a function of the number of training tokens. A gradual improvement was observed with an increase in the number of training tokens. A peak recognition rate of 42.50 percent was obtained for the network trained with 18 tokens.

Evaluation Results

The Introvoice was trained with the same set of 20 words, and the number of training tokens was varied from 6 to 18. A peak recognition rate of 37.5 percent was obtained when the system was trained with 15 tokens. The recognition rate did not show a steady improvement as a function of number of tokens.

An average intelligibility of 42.38 percent was scored for the subject's speech by the five listeners. The results of the Rhyme test are summarized in **Table 3**. Most of the errors were associated with the phonemes that required extreme articulatory positions (i.e., stops like /t/, /d/, /p/ and fricatives). Again, studying these patterns is the focus of our current research.

CONCLUSION

This study presented a neural network approach to recognizing isolated words spoken by a dysarthric speaker. The network ability to recognize the target words was com-

Table 2.
Confusion matrix for FFT network trained with 18 tokens.

	1	2	3	4	5	6	7	8	9	10	11	12	13	14	15	16	17	18	19	20
1	3						1													
2		4																		
3			3																	
4				3	1															
5					3															
6						4														
7							4													
8								3				1								
9									4											
10										4										
11	1					1					0								1	1
12		1										1	2							
13			1										3							
14														4						
15		1													3					
16																3			1	
17									1								3			
18						2												2		
19																			4	
20					1															3

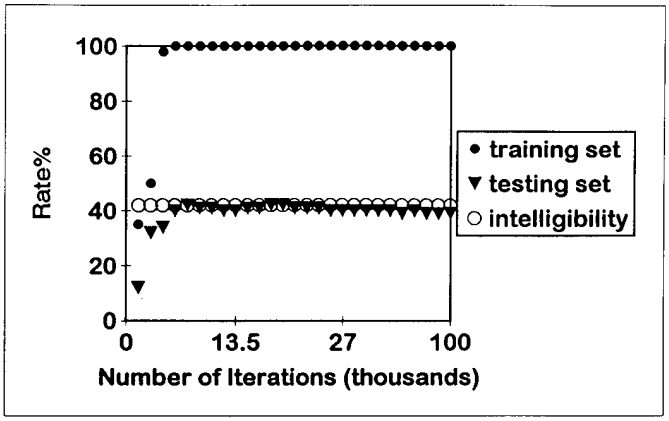

Figure 4.
Recognition results for formant network.

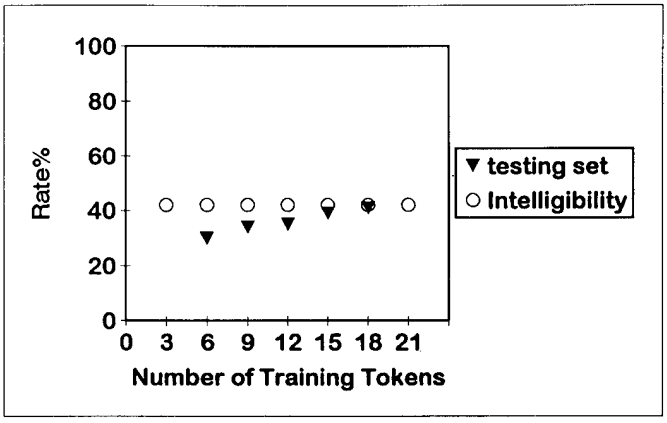

Figure 5.
Effect of training for formant network.

pared with that of the Introvoice speech recognition system and the intelligibility rating for a speaker determined by experienced human listeners. The results show the ability of the developed networks to successfully recognize dysarthric speech despite its large variability. These networks clearly outperformed both the human listeners and the Introvoice commercial system. These results are summarized in **Figure 6**.

The results also demonstrated that an increase in recognition rate was observed with addition of the energy level to the input feature vector. Adding more features such as zero-crossing rate to the input vector will possibly further improve the recognition performance. Currently, we are looking into those features that are most related to the intelligibility of dysarthric speech. More research is also needed to establish the validity of the approach under

Table 3.
Confusion matrix for Rhyme Test.

	p	t	k	f	s	h	m	n	b	d	g	w	r	l	dz	v	z	j
p	20				1	1			22			1						
t	33	11	7						20	2	1	1						
k		2	36		1	1				1	2	1						1
f	5	5		17	4	8			5									
s	2	10	1	1	38	15	1		5	9	1							6
h						35						1	9					
m					3	4	51	1					1					
n					3		5	18		3		1						
b	17	3			5	2			30			3						
d		8	5			2	3	2	2	10	4	2	4	2				
g	2		13							1	14							
w	4			3	1	13			4			18	15			1		1
r	1	1	1		1	16	1	1				6	11	3				3
l	2	1			3	10	1	1	1	1	1	4	11	9	3			12
dz							1		10									4
v																		
z																		
j																		

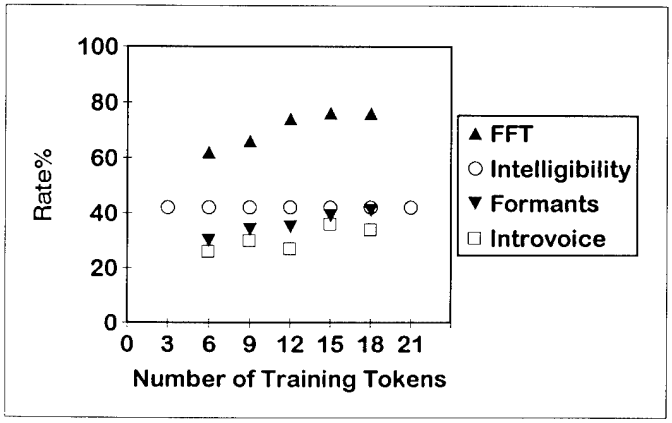

Figure 6.
Summary of recognition results.

greater phonemic environment, expanded vocabulary, and with a group of speakers.

We would like to emphasize that the use of a single case with dysarthria as a result of cerebral palsy is appropriate for this study. Cerebral palsy is the most common cause of dysarthria. Variability is the hallmark of this disorder and poses the challenge to speech recognition technology. Although the study focuses on one diagnosis, many of the features of cerebral palsy speech (e.g., variability) are also features of dysarthria that is the result of traumatic brain injury, stroke, or multiple sclerosis. The data presented here, therefore, have implications for dysarthric individuals other than those with cerebral palsy.

We believe that the approach described in this study is an important step toward automatic recognition of dysarthric speech. This will eventually lead to the development of effective voice-input communication and control assistive devices for individuals with cerebral palsy and others with neurogenic communication disorders.

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REFERENCES

1. Elliot D. A computerized speech recognizer for dysarthric speech. In: Proceedings of the 40th Annual Conference on Engineering in Medicine and Biology, Niagara Falls, NY, 1987:9:63.
2. Miller GE, Etter BD, Bartholomew JC. Analysis of voice processing for the control of devices to aid the disabled. In: Proceedings of the 12th Annual RESNA Conference, 1989, New Orleans, LA. Washington, DC: RESNA Press, 1989:410–2.
3. Lee WC, Blackstone SW, Pook GK. Dysarthric speech input to expert systems: electronic mail and daily job activities. In: Proceedings of the American Voice Input/Output Society, 1987, San Jose, CA: AVIOS 1987:33–43.
4. Carlson GS, Bernstein G. Speech recognition of impaired speech. In: Proceedings of the 10th Annual RESNA Conference, 1987, San Jose, CA. Washington, DC: RESNA Press, 1987:103–5.
5. Goodenough C, Rosen M. Towards a method for computer interface design using speech recognition. In: Proceedings of the 14th Annual RESNA Conference, 1991, Kansas City, MO. Washington, DC: RESNA Press 1991:328–9.
6. Sy BK, Horowitz DM. A statistical causal model for assessment of dysarthric speech and the utility of computer based speech recognition. IEEE Trans Biomed Eng 1993:40(12):1282–98.
7. Coleman CL, Meyers LS. Computer recognition of the speech of adults with cerebral palsy and dysarthria. Augment Alternat Commun 1991:7(1):34–42.
8. Deller JR, Jr, Hsu D, Ferrier LJ. The use of hidden Markov modeling for recognition of dysarthric speech. Comput Methods Programs Biomed (Netherlands) 1991:35(2):125–39.
9. Boonzaier DA, Limon A. Dysarthric speech recognition: a hidden Markov modeling approach. In: Proceedings of the 16th Annual RESNA Conference, 1993, Las Vegas, NV. Washington, DC: RESNA Press, 1993:108–10.
10. Lippmann RP. Review of neural networks for speech recognition. Neural Computat 1989:1:1–38.
11. Hecht-Nielsen R. Neurocomputing. New York: Addison-Wesley Publishing Co., 1990.
12. Bottou L, Soulie FF. Speaker-independent isolated digit recognition: multilayer perceptrons vs. dynamic time warping. Neural Networks 1990:3:453–65.
13. Lerner S, Deller J. Neural network learning of spectral features of non-verbal speech. In: Proceedings of the IEEE International Conference on Acoustics, Speech, and Signal Processing, New York, NY, 1988:24:43–6.
14. Mysak ED. Cerebral Palsy. In: Shanes GH, Wiig EH, eds. Human communication disorders. Columbus, OH: Merill Publishing Co., 1986: 513–60.
15. Lehiste I. Some acoustic characteristics of dysarthric speech. Basil, Switzerland: S. Karger, 1965.
16. ILS [Interactive Laboratory System] reference manuals, STI International, Santa Barbara, CA: STI Neural Networks, 1990.
17. Tompkins WJ, Webster JG. Design of microcomputer based medical instrumentation. Englewood Cliffs, NJ: Prentice Hall, Inc., 1981.
18. O'Shaughnessy D. Speech communication: human and machine. New York: Addison-Wesley, 1986.
19. Kammerer B, Kupper W. Experiments for isolated-word recognition using single and two-layer perceptrons. Neural Networks 1990:3:693–706.
20. Professional II Plus Software Manuals, Neural Ware, 1990.
21. House A, Williams C, Hecker M, Kryter K. Articulatory testing methods: consonant differentiation in a closed response set. J Acoust Soc Am 1965:37:158–66.
22. Fairbanks G. Test of phonemic differentiation: the rhyme test. J Acoust Soc Am 1958:30:7:596–600.
23. ANSI S3.2. American standard method for measuring the intelligibility of speech over communication systems. New York: Acoustical Society of America, 1989.

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CLINICAL REPORT

Energy Cost and Locomotive Economy of Handbike and Rowcycle Propulsion by Persons with Spinal Cord Injury

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Abstract—Seven subjects with chronic paralysis due to spinal cord injury completed a series of experiments to 1) determine and compare the metabolic cost of propelling the Handbike and Rowcycle, and 2) evaluate the potential of these upper body–powered devices for improving the cardiorespiratory fitness of persons with lower limb disabilities. Mean intrasubject differences between the Handbike and Rowcycle rides for heart rate, minute ventilation, oxygen uptake, and net locomotive energy cost were small and did not reach statistical significance for any of the ride conditions. Lower net locomotive energy cost (greater economy) during a 5.5 mi·hr^{-1} ride condition predicted vehicle preference in all cases (P=0.008). The range of values for percent peak oxygen uptake suggests that all but one of the subjects were able to utilize either vehicle at an intensity sufficient for improving and maintaining cardiorespiratory fitness without undue fatigue.

Key words: *exertion, oxygen consumption, spinal cord injury, upper body aerobic exercise.*

INTRODUCTION

Exercise training and regular physical activity may improve the health and quality of life of persons with spinal cord injury (SCI) by slowing or reversing physiological degeneration, favorably altering cardiovascular disease and diabetes risk factors, improving psychological status and independence, and allowing the performance of daily activities with less fatigue (1–3). Recently, efforts have been intensified to develop and market exercise devices for persons with lower limb disabilities (4,5). To be useful for the population with lower limb disabilities, the inexperienced operator must be able to use a device without undue fatigue. Excessive metabolic stress is likely to discourage the use of the equipment for exercise that is of adequate intensity and duration to improve and maintain cardiorespiratory fitness (3). In addition, the more skilled and/or fit individual should be able to operate the device at an intensity that is sufficient to promote cardiorespiratory fitness and satisfy recreational objectives.

The present investigation was designed to: 1) determine and compare the metabolic cost, to novice riders, of propelling two upper body–powered vehicles, the Handbike and the Rowcycle, which are suitable for use by persons with lower limb paralysis due to SCI, and 2) evaluate the potential of these devices for improving the cardiorespiratory fitness of these individuals.

METHODS

The Handbike and Rowcycle

The Handbike (**Figure 1a**) is an upper body–powered vehicle which was developed at the Department of Veterans Affairs Rehabilitation Research and Development Center, Palo Alto, CA. It is an extension of an earlier model known

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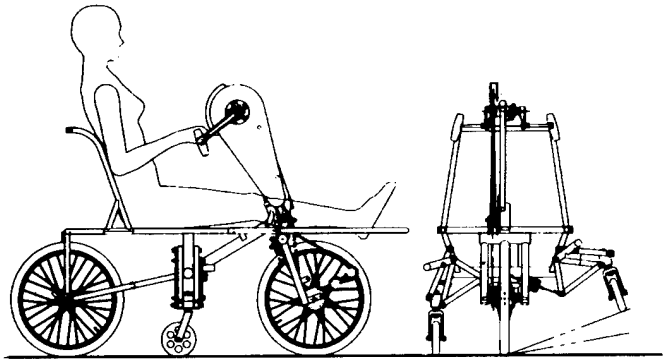

Figure 1a.
The Handbike.

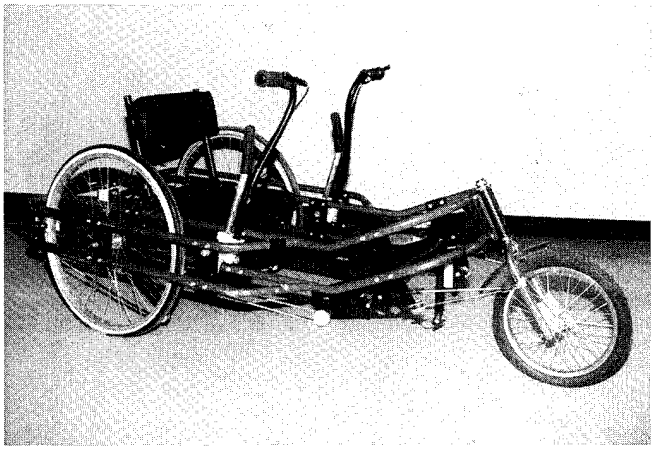

Figure 1b.
The Rowcycle.

as the Para-bike (6–8). The present investigation was conducted as part of a Department of Veterans Affairs Technology Transfer Service evaluation of the Handbike.

The Handbike is powered by a synchronous arm-cranking motion. The moveable crank tower is connected to the front fork which allows steering movements without interruption of cranking. Adjustable side wheels may be kept in contact with the riding surface or lifted so that the vehicle can be balanced on two wheels while riding. In the lifted position, the side wheels will touch the ground at a level that will prevent the rider from tipping over during a sharp turn or after stopping. The gearing system consists of a five-speed Sturmey-Archer hub and a two-speed derailleur which shifts the drivechain from one size hub sprocket to another. The result is 10 possible gearing combinations. Although the inventor suggests the possibility that there is some overlap between gears in the larger and smaller sprockets, he does not specify gear- or drive-ratios (7). The Handbike has an 85 cm wheel base, is 63.5 cm wide with the side wheels in the down position, and weighs 21 kg. It is commercially available from New Dimensions Design, Elmira, OR.

The Rowcycle (**Figure 1b**) used for this study was purchased from Rowcycle Company, Fresno, CA (5,9). It is a three-wheeled vehicle powered by a rowing motion (synchronous or asynchronous). This vehicle was chosen for comparison because it is propelled using an upper body motion that is biomechanically different from that used when riding the Handbike. Steering is accomplished by tilting to the left or right in the seat. The Rowcycle has a three-speed, adjustable leverage transmission which operates by altering the row-handle fulcrum. Gearing, powering, and braking are independent for each side of the Rowcycle. The Rowcycle is 183 cm in length, has a width of 90 cm, and weighs 30 kg.

Subjects

Seven apparently healthy individuals with SCI, six male and one female, participated in this study. Characteristics of the participants are shown in **Table 1**. The study procedures were approved by the Human Subjects Subcommittee of Edward Hines Jr., Department of Veterans Affairs Hospital (Hines VAH). Informed consent was obtained from all subjects as well as a statement from each subject's personal physician giving approval for participation in the study. All subjects were required to have had a physical examination in the 12 months prior to the study and to fill out a Medical History Questionnaire. The Medical History Questionnaire was used in the initial screening of volunteers and to identify contraindications to exercise testing. Data collection was carried out in the Rehabilitation Research and Development (Rehab R&D) Center's Physical Performance Research Laboratory and on the Hines VAH grounds.

Wheelchair Graded Exercise Tests

All subjects were given a maximal wheelchair graded exercise test using the Wheelchair Aerobic Fitness Trainer (WAFT), a wheelchair ergometer developed at the Hines VAH Rehab R&D Center (10–12). A progressive exercise protocol was utilized in which the initial workload was set at 6 watts and increased by ~7 watts every 3 minutes until volitional exhaustion, or until the subject was unable to maintain the workload (12). Testing procedures conformed

Table 1.
Peak values during maximal effort wheelchair ergometry.

Subject	Sex	Age yr	Level of Injury[a]	Body Wt. kg	$\dot{V}O_2$ L·min^{-1}	HR b·min^{-1}	$\dot{V}_E$ L·min^{-1}
1	M	36	T_{12}-L_1[c]	81.0	1.76	200	108.8
2	M	34	T_{5-6}[c]	68.0	1.35	196	65.2
3	M	25	C_{6-7}[inc]	94.8	1.40	120	55.8
4	M	32	T_{5-6}[c]	70.3	1.60	176	56.3
5	M	40	T_{6-7}[c]	59.0	1.11	173	64.7
6	M	29	T_{10-11}[inc]	72.6	2.09	192	93.4
7	F	53	T_7[inc]	48.5	0.80	150	33.3

[a]C = cervical; T = thoracic; L = lumbar; [c] = complete injury, [inc] = imcomplete injury

to guidelines established by the American College of Sports Medicine (13).

Heart rate (HR) was derived from a standard electrocardiogram (ECG). A Q3000B Stress Test Monitor (Quinton Instrument Company, Seattle, WA) was used for visual monitoring and recording of the ECG. Blood pressures were measured by auscultation at rest and in recovery, as well as during short pauses (<30 seconds) between stages. Expired gases were collected and analyzed by an open circuit method using the Horizon Advanced Exercise, Metabolic Measurement Cart (MMC), SensorMedics Corporation, Yorba Linda, CA. Samples of expired gases were analyzed for concentrations of carbon dioxide using a nondispersive infrared technique and concentrations of oxygen using a polarographic sensor cell. Prior to and after each test, analyzer calibration was checked using reference gases and room air.

Gas collection was begun in the last minute of a 15-minute rest period preceding the exercise tests and continued throughout the test and at least 2 minutes into recovery. Ratings of perceived exertion (RPE) were obtained by using Borg's 15-point graded category scale (13). Ratings were taken during the last 30 seconds of each 3-minute stage and the final 30 seconds of the peak workload.

Onset of blood lactate accumulation by gas exchange ($OBLA_{ge}$) was derived from MMC Horizon computer plots. Analysis of the plotted data from the maximal wheelchair graded exercise test was based upon criteria proposed by Wasserman and McIlroy (14). Data points were plotted at 15-second intervals. Two investigators (KCM and WEL) conducted independent analysis of the plots. These analyses were compared and, when they differed, the factors that influenced the determination of the $OBLA_{ge}$ were discussed by the investigators, and a single value decided upon.

Handbike and Rowcycle Training and Practice

Subjects were randomly assigned an order for Handbike and Rowcycle training and testing. Each subject was then given a supervised training and practice period, lasting approximately 60 minutes, on the first of the two vehicles. Before data collection was started, each subject demonstrated the ability to safely stop, steer, execute 90° and 180° turns, and propel the vehicle at different speeds. When the subject indicated that he or she was ready and the investigators were satisfied with the subject's skill level, data collection procedures were initiated. These same steps were repeated on a different day using the other experimental vehicle, or on the same day following a rest period of at least 90 minutes.

Data Collection for Handbike and Rowcycle Field Tests

Subjects performed two rides each on the Handbike and Rowcycle; one at 5.5 mi·hr^{-1} and the other at a freely chosen speed (FCS). In addition, if the subject felt he or she could tolerate it, a third ride was performed on each device at 120 percent of the FCS. Assignment to initial testing device (Handbike or Rowcycle) and ride condition (5.5 mi·hr^{-1} vs. FCS) were made by randomization. The rides were 5–7 minutes in length. A 140-inch drive ratio was used for both vehicles during data collection rides. All rides were performed in the Hines VAH parking lot on a 0.3 mile (0.48 km), curved course which required two gradual 180° turns during the period of expired gas collection.

Ride speed was monitored by an outrider riding next to each subject on a bicycle equipped with a Cateye Cyclocomputer (Performance Bicycle Shop, Chapel Hill, NC). Instructions to increase or decrease speed were given as necessary. For the FCS condition, the subject chose a

speed during the first 1 to 2 minutes and was asked to maintain that speed during the remainder of the ride. Mean ride speed was calculated by dividing the total distance traveled (recorded from the Cateye Cyclocomputer) by the total time elapsed during the ride. Heart rate was monitored using a CIC Wireless Heartwatch (Performance Bicycle Shop) with a telemetry display mounted on subjects' headgear. The Heartwatches used in the study were checked for accuracy prior to testing using an ECG readout from the Quinton 3000 ECG. The outrider reported elapsed ride time, speed, and heart rate into a portable cassette tape recorder during each ride. Total ride time and distance were recorded from the cyclocomputer at the end of each ride.

Expired gases were collected using an open circuit method during the final 45 to 90 seconds of each ride. Subjects were equipped with a rubber mouthpiece, Hans Rudolph two-way valve, and nose clip. Expired gases flowed through a 7-ft hose connected to a three-way directional stopcock. During the collection period, the stopcock was adjusted so that the expired gases were collected in a 100 L neoprene meteorological balloon that was suspended inside a 35 gal plastic container. During testing, the plastic container was carried on the front of another bicycle being ridden by an investigator as shown in **Figure 2.** Gas samples were analyzed within 15 minutes after collection using the Horizon Advanced Exercise MMC operated in the manual mode.

During the last 15 seconds of the final minute of each test ride, the outrider asked the subject to choose an RPE (13). The subject then glanced at a copy of the RPE scale which was tied to his or her leg. The rating was reported at the end of each ride.

After a subject had been tested on both devices, he or she was asked to fill out a four-part questionnaire with questions regarding transferring and seating, vehicle operation, overall impressions, and vehicle preference.

Gross energy cost (GEC) for each riding condition and resting energy expenditure were calculated using respiratory exchange ratio and oxygen uptake ($\dot{V}O_2$) (15). Resting energy expenditure was calculated using data collected during the final 1 to 2 minutes of a 15-minute rest period preceding the wheelchair graded exercise test.

Net locomotive energy cost (NLEC), or net kcal per unit of body weight per unit of distance traveled, was calculated as proposed by Glaser et al. (3,16), for all experimental conditions:

$$NLEC = (GEC - E)(Wt - D)^{-1} \quad [1]$$

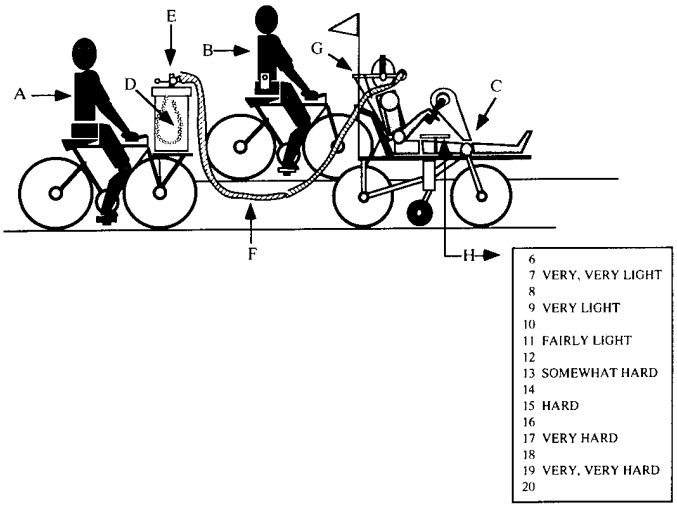

Figure 2.
Illustration of the field testing method employed for data collection during Handbike and Rowcycle propulsion: (A) investigator riding bicycle with gas collection bag; (B) outrider/pacer wearing tape recorder for heart rate, speed, time, distance, and rating of perceived exertion storage; (C) subject riding vehicle (Handbike pictured); (D) 100 L neoprene meteorological balloon suspended inside a 35 gal plastic container; (E) three-way directional stopcock; (F) 7 ft large bore hose; (G) Heartwatch display, Hans-Rudolph non-rebreathing valve, head support for valve, and suspension support for hose; (H) rating of perceived exertion scale attached to subject's leg.

where GEC is the gross energy cost in kcal, E is the resting energy expenditure in kcal, Wt is the subject's weight in kg, and D is the distance traveled in km (16). The units of NLEC are $kcal \cdot kg^{-1} \cdot km^{-1}$. As stated by Glaser and colleagues, "Since gross energy cost is expressed in kcal per unit of time, it provides little information as to the metabolic cost for an individual to travel a given distance. NLEC, however, provides an index to express the relative energy cost (corrected for resting metabolism and body weight) per km of distance traveled. Therefore, to facilitate comparison, we propose that this index be used to express the efficiency [economy] of various locomotive tasks" (16).

Statistical Analysis

The physiologic responses of subjects during Handbike and Rowcycle propulsion were compared using two-way analysis of variance for repeated measures. Dependent variables included ride speed, HR, $\dot{V}O_2$, minute ventilation ($\dot{V}_E$), and NLEC and independent variables were ride condition (5.5 $mi \cdot h^{-1}$, FCS and 120 percent FCS) and vehicle (Handbike vs. Rowcycle).

Where appropriate, paired *t*-tests were used for *post hoc* comparisons without correction for multiple comparisons. Least squares linear regression was also employed to examine the relationships between ride speed, $\dot{V}O_2$, and NLEC. A binomial probability formula was used to determine whether a statistically significant relationship existed between vehicle preference and NLEC at the 5.5 $mi \cdot h^{-1}$ ride condition (17) and single sample *t*-tests were performed on the mean slopes relating ride speed to NLEC for each vehicle. Analyses were performed on a Macintosh LC personal computer using the StatView 4.0 Statistical Analysis Package (Abacus Concepts, Calabasas, CA). An alpha level of 0.05 was used to denote statistical significance.

RESULTS

Physiologic Responses

Mean values for the measured or calculated parameters are presented in **Tables 2–4.** Within subject differences between the Handbike and Rowcycle for speed, HR, $\dot{V}_E$, $\dot{V}O_2$, and NLEC did not reach statistical significance for any of the three ride conditions. Ride speed was significantly different between each of the three conditions for both the Handbike and Rowcycle ($P<0.05$ for all comparisons). NLEC did not differ significantly between ride conditions for either vehicle. In order to examine whether a relationship existed between ride speed

Table 2.
Mean individual values for speed and physiologic variables during Handbike and Rowcycle rides in the 5.5 $mi \cdot hr^{-1}$ condition.

	Speed ($mi \cdot hr^{-1}$)		$\dot{V}O_2$ ($L \cdot min^{-1}$)		$\dot{V}_E$ ($L \cdot min^{-1}$)		HR ($b \cdot min^{-1}$)		NLEC ($kcal \cdot kg^{-1} \cdot km^{-1}$)	
Subject	**HB**	**RC**	**HB**	**RC**	**HB**	**RC**	**HB**	**RC**	**HB**	**RC**
1	5.9	5.9	0.782	0.832	22.88	24.19	128	113	0.248	0.269
2	5.9	5.8	0.816	0.659	26.84	27.08	132	137	0.246	0.267
3	5.9	5.6	0.802	0.712	39.38	40.19	94	97	0.206	0.188
4	5.6	5.8	0.496	0.486	22.14	19.38	116	99	0.139	0.124
5	5.6	5.2	0.616	0.563	24.64	22.99	155	134	0.290	0.263
6	5.6	5.5	0.571	0.694	16.63	21.47	93	128	0.122	0.177
7	5.9	5.0	0.701	0.450	26.23	21.40	116	113	0.545	0.383
Mean	5.8	5.5	0.683	0.628	25.53	25.24	119	117	0.257	0.239
SD	0.2	0.3	0.125	0.136	6.98	7.03	22	16	0.141	0.084

$\dot{V}O_2$ = oxygen uptake, $\dot{V}_E$ = minute ventillation, HB = Handbike, RC = Rowcycle, HR = heart rate, NLEC = net locomotive energy cost

Table 3.
Mean individual values for speed and physiologic variables during Handbike and Rowcycle rides in the freely chosen speed condition.

	Speed ($mi \cdot hr^{-1}$)		$\dot{V}O_2$ ($L \cdot min^{-1}$)		$\dot{V}_E$ ($L \cdot min^{-1}$)		HR ($b \cdot min^{-1}$)		NLEC ($kcal \cdot kg^{-1} \cdot km^{-1}$)	
Subject	**HB**	**RC**	**HB**	**RC**	**HB**	**RC**	**HB**	**RC**	**HB**	**RC**
2	6.4	7.1	0.648	0.693	30.80	30.11	163	148	0.226	0.230
3	6.0	6.2	0.773	0.899	53.24	51.37	128	113	0.195	0.240
4	7.0	7.0	0.679	0.760	36.97	34.71	137	113	0.194	0.235
5	6.7	7.4	1.104	0.732	44.86	32.54	159	153	0.509	0.278
6	6.6	9.8	0.987	1.706	33.56	108.70	121	184	0.300	0.464
7	5.9	6.0	0.703	0.479	29.06	27.39	125	122	0.533	0.359
Mean	6.4	7.3	0.816	0.881	38.08	47.47	134	136	0.326	0.301
SD	0.4	1.4	0.186	0.424	9.29	31.16	25	32	0.156	0.093

$\dot{V}O_2$ = oxygen uptake, $\dot{V}_E$ = minute ventillation, HB = Handbike, RC = Rowcycle, HR = heart rate, NLEC = net locomotive energy cost

Table 4.
Mean individual values for speed and physiologic variables during Handbike and Rowcycle rides in the 120% freely chosen speed condition.

	Speed ($mi \cdot hr^{-1}$)		$\dot{V}O_2$ ($L \cdot min^{-1}$)		$\dot{V}_E$ ($L \cdot min^{-1}$)		HR ($b \cdot min^{-1}$)		NLEC ($kcal \cdot kg^{-1} \cdot km^{-1}$)	
Subject	HB	RC	HB	RC	HB	RC	HB	RC	HB	RC
2	9.0	9.0	0.954	1.020	53.27	57.71	189	183	0.289	0.310
3	•	6.4	•	0.959	•	72.04	•	112	•	0.255
4	9.4	9.1	0.822	1.069	48.27	59.17	151	152	0.198	0.297
5	7.8	8.7	1.369	0.975	57.84	45.51	166	158	0.572	0.340
6	9.4	11.5	1.375	1.725	43.43	112.87	143	186	0.344	0.386
Mean	8.9	8.9	1.130	1.150	51.20	69.46	162	158	0.351	0.318
SD	0.8	1.8	0.290	0.320	7.02	26.02	20	30	0.159	0.049

$\dot{V}O_2$ = oxygen uptake, $\dot{V}_E$ = minute ventillation, HB = Handbike, RC = Rowcycle, HR = heart rate, NLEC = net locomotive energy cost

and NLEC, the slope of the regression line of NLEC ($kcal \cdot kg^{-1} \cdot km^{-1}$) on ride speed ($mi \cdot h^{-1}$) was calculated for each subject for Handbike and Rowcycle rides. A tendency was noted for a slight increase in NLEC with increasing speed (meanβ=0.042±0.099 $kcal \cdot kg^{-1} \cdot km^{-1}$ for the Handbike and 0.031 ±0.036 $kcal \cdot kg^{-1} \cdot km^{-1}$ for the Rowcycle), but this was not statistically different from zero for either vehicle.

Target Training Intensities for Oxygen Uptake

One objective of this study was to evaluate the utility of these devices for exercise training to improve cardiorespiratory fitness. The American College of Sports Medicine has published guidelines for prescribing exercise intensity for this purpose (13). They suggest that the training-sensitive range is from 40–85 percent of $\dot{V}O_{2peak}$. The mean percent of $\dot{V}O_{2peak}$ was within this range for all ride conditions with both vehicles. For the 5.5 $mi \cdot h^{-1}$ condition, percent $\dot{V}O_{2peak}$ for the Handbike and Rowcycle were 52.0 ±20.4 percent and 45.4±9.7 percent, respectively. The corresponding values were 63.5±24.2 percent and 62.2±12.1 percent for the FCS condition and 78.0±31.6 percent and 76.4±9 percent in the 120 percent of FCS condition. Furthermore, of the six persons who completed at least two rides on each vehicle, all six were within the training-sensitive range on at least one ride with the Rowcycle and five of the six were within the training-sensitive range on at least one ride with the Handbike. Percent $\dot{V}O_{2peak}$ was above the training-sensitive range for both the 5.5 $mi \cdot h^{-1}$ and FCS conditions on the Handbike for one individual (88 percent for both rides by subject no. 7).

$OBLA_{ge}$ During Wheelchair Ergometry and Its Relationship To Oxygen Uptake and RPE During Handbike and Rowcycle Rides

Average $OBLA_{ge}$ during the maximal effort wheelchair ergometry tests occurred 57.0±5.0 percent of $\dot{V}O_{2peak}$ and RPEs of 12 to 13. ($OBLA_{ge}$ could not be determined for one subject.) The $OBLA_{ge}$ during maximal effort wheelchair ergometry may or may not be similar to that measured during Handbike or Rowcycle propulsion. Recognizing that there are limitations to the usefulness of any interpretations made from these data, we felt that it was worth considering from a qualitative perspective. The number of subjects exceeding $OBLA_{ge}$ from their wheelchair tests within each condition was one of six for the Handbike and zero of six for the Rowcycle in the 5.5 $mi \cdot h^{-1}$ condition, two of five for the Handbike and three of five for the Rowcycle at FCS, and three of four for the Handbike and four of four for the Rowcycle in the 120 percent FCS condition.

Vehicle Preference and Relationship to NLEC

In response to a questionnaire, four of the seven subjects reported that they preferred the Handbike, while three preferred the Rowcycle. A binomial probability formula was applied to determine whether there was a statistically significant relationship between vehicle preference and NLEC under the 5.5 $mi \cdot h^{-1}$ condition (i.e., whether the vehicle which elicited the lower NLEC was consistently preferred). This ride condition was chosen because it was the only one for which speed was held constant between subjects. Lower NLEC predicted vehicle preference in all seven cases, which was highly significant (p=0.008).

Relationship Between Oxygen Uptake and Speed of Rowcycle and Handbike Propulsion

Nonsignificant correlations ($p>0.05$) were found between ride speed during Handbike propulsion—all conditions combined—and oxygen uptake in $L \cdot min^{-1}$ ($r=0.65$) and $mL \cdot kg \cdot min^{-1}$ ($r=0.51$) (**Figure 3**). Stronger and statistically significant relationships ($p<0.01$) were found for Rowcycle propulsion, $r=0.88$ and $r=0.93$ for speed versus oxygen uptake in $L \cdot min^{-1}$ and $mL \cdot min^{-1} \cdot kg^{-1}$, respectively (**Figure 4**).

Comparison With the Medrano Study

A similar energy cost study was performed with the Rowcycle by Medrano (18). Oxygen consumption was measured in 10 subjects with lower limb disability while propelling the Rowcycle at 4, 6, and 8 $mi \cdot hr^{-1}$. Gear(s) and/or drive ratio(s) utilized were not specified. Raw data were provided for the 8 $mi \cdot hr^{-1}$. Analysis of our data and the raw data of Medrano showed that the regression lines relating speed and oxygen uptake (in $mL \cdot kg \cdot min^{-1}$) were nearly identical, and did not differ significantly in slope or intercept (data not shown).

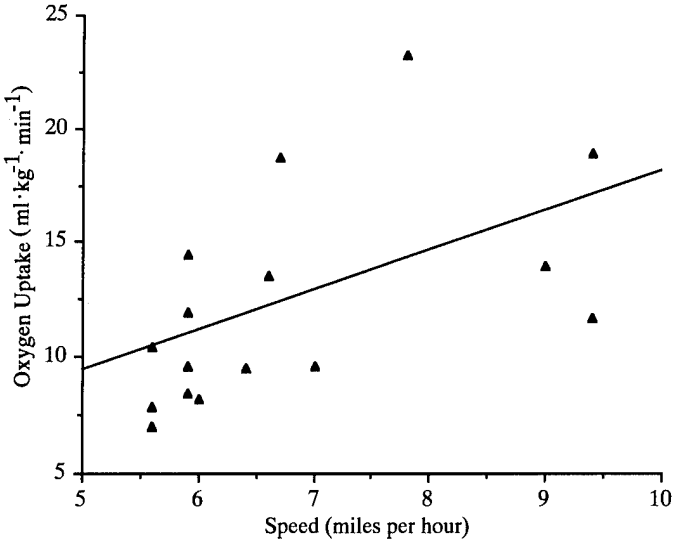

Figure 3.
Oxygen uptake ($ml \cdot kg^{-1} \cdot min^{-1}$) vs. Handbike speed ($mi \cdot hr^{-1}$), $r=0.51$, $P>0.05$, oxygen uptake = 0.79+1.73 (speed).

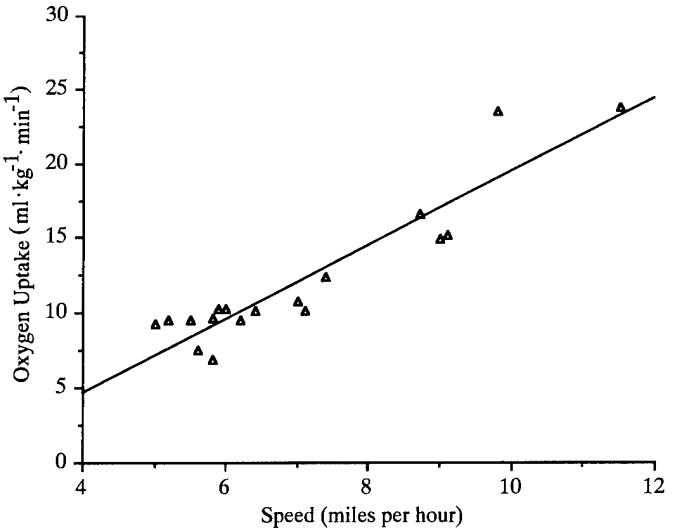

Figure 4.
Oxygen uptake ($ml \cdot kg^{-1} \cdot min^{-1}$) vs. Rowcycle speed ($mi \cdot hr^{-1}$), $r=0.93$, $P<0.01$, oxygen uptake = -5.17+2.46 (speed).

DISCUSSION

Physiological Responses

Factors which influence the magnitude of physiological responses to wheelchair locomotion, as summarized by Glaser et al. (16), include: the fitness level of the user, characteristics of the wheelchair used, velocity of locomotion, and architectural conditions (floor or ground surface, grade, etc.). These same determinants also apply to propulsion of other upper body–powered vehicles. The present investigation assessed the physiological responses to propelling the Handbike and Rowcycle, which utilize different methods of propulsion (synchronous arm-cranking vs. rowing), at similar velocities and drive ratios. All of the subjects were persons with chronic SCI who were inexperienced Handbike and Rowcycle riders. Tests were done out-of-doors to simulate, as closely as possible, environmental conditions encountered during normal use of these vehicles. Within subject differences in HR, $\dot{V}O_2$, $\dot{V}_E$, or NLEC taken during propulsion of the Handbike and Rowcycle under the three experimental conditions (5.5 $mi \cdot hr^{-1}$, FCS, and 120 percent of FCS) were small and not statistically significant.

Comparisons between the Handbike and Rowcycle were limited to riding conditions likely to be encountered by the novice rider. This strategy was employed because early experiences would be expected to exert the greatest influence on the likelihood that an individual would continue to use either vehicle for fitness training and/or recreation. The Handbike and Rowcycle were each tested in only one drive ratio. The 140-inch gear was chosen be-

cause it was common to both the Handbike and Rowcycle. The Handbike was tested only with the side wheels in the down position, which would be used by the novice rider. The time necessary to develop the requisite skill to ride the Handbike on two wheels, as well as concerns for rider safety during expired gas collection, prevented testing of subjects with the side wheels raised.

Potential For Aerobic Conditioning Using The Handbike and Rowcycle

The ACSM recommends a training intensity of 40 to 85 percent $\dot{V}O_{2peak}$ for aerobic conditioning (13). One objective of the present investigation was to evaluate the potential for aerobic conditioning through regular use of the Handbike and Rowcycle by the inexperienced rider with SCI. Of the six subjects who completed rides in at least two conditions, all were within the training-sensitive range (40-85 percent $\dot{V}O_{2peak}$) for at least one ride on the Rowcycle. Five of six subjects were within this range during at least one Handbike ride. One subject was slightly above the training range on both the 5.5 mi·hr^{-1} and FCS Handbike rides (88 percent $\dot{V}O_{2peak}$ for both). Therefore, it appears that most riders with paraplegia should be able to utilize either vehicle for exercise conditioning. The current recommendations of the ACSM for apparently healthy adults (13) were used as a guide in this investigation, but a study by Hooker and Wells (19) suggests that the ACSM guidelines may not be appropriate for improving the fitness of persons with SCI. Further investigation is needed to clarify the optimal training intensity range for aerobic conditioning in this group.

$OBLA_{ge}$ was determined as an additional point of comparison. During the 5.5 mi·hr^{-1} condition, all but one subject was below $OBLA_{ge}$ for the Handbike and none were above $OBLA_{ge}$ during Rowcycle propulsion in this condition. In contrast, all subjects were above $OBLA_{ge}$ for the Rowcycle ride at the 120 percent FCS condition and all but one rider was above $OBLA_{ge}$ during the corresponding Handbike rides. This further suggests that both vehicles may be used for exercise of an intensity covering a wide enough range to effectively train the cardiorespiratory system of persons of varying degrees of physical fitness.

Vehicle Preference

Analysis of the data indicates that locomotive economy, as indicated by NLEC at 5.5 mi·hr^{-1}, was significantly related to the subjects' vehicle preference ($p=0.008$). The vehicle associated with the lower NLEC predicted subjects' vehicle preference in all seven cases, even though no overall trend was observed for preference of one vehicle over the other (four preferred the Rowcycle and three the Handbike). Therefore, locomotive economy appears to be an important determinant of vehicle preference. This finding has clinical implications because patient compliance with a prescription for a mobility device, whether for activities of daily living, recreation, sports, or aerobic conditioning, will be significantly impacted by the perception of the physical and psychological stress experienced when using the device.

Glaser et al. (16) compared the energy cost and cardiopulmonary responses to wheelchair locomotion on tile and carpet in 9 wheelchair-dependent men. Mean NLEC for wheelchair propulsion at 3.0 km·hr^{-1} (1.34 mi·hr^{-1} was 0.46±0.03 and 0.80±0.06 kcal·kg^{-1}·km^{-1} for tile and carpet, respectively. Mean NLEC values were considerably lower for the Handbike and Rowcycle (0.24–0.35 kcal·kg^{-1}·km^{-1}), indicating that there was greater relative locomotive economy associated with propelling the Handbike and Rowcycle on a paved surface.

Relationship Between Speed and $\dot{V}O_2$ For The Handbike and Rowcycle

The relationship between speed and $\dot{V}O_2$ was stronger for the Rowcycle than for the Handbike. One possible explanation for this may be that riding the Handbike requires greater motor skill (balance, coordination) than riding the Rowcycle, resulting in a more variable speed-$\dot{V}O_2$ relationship. Since all of the riders were inexperienced, it must be remembered that differences in skill requirements between the two vehicles may not affect rider performance in the same way following a more extended period of practice and skill development.

Comparison of the data from the present study with that from Medrano (18) shows that the two studies yielded similar results with regard to the energy cost of propelling the Rowcycle. Combining the data from the two studies results in a strong correlation between speed and $\dot{V}O_2$ in mL·min^{-1} ($r=0.94$, $p<0.001$). The resulting regression equation, derived from gas analysis of 27 Rowcycle rides was:

$$\dot{V}O_2 = 2.46\ (\text{speed}) - 5.09 \qquad [2]$$

where $\dot{V}O_2$ is in mL·min^{-1}·kg^{-1} and speed of propulsion is in mi·hr^{-1}. This may be useful for exercise prescription and/or field testing of persons with lower limb disabilities to predict oxygen uptake from a known speed.

Handbike and Rowcycle Use In An Individual With a High Level Spinal Lesion

One of the research subjects found it very difficult to ride the Handbike, but rode the Rowcycle with relative ease. This individual had an incomplete spinal cord lesion at the C_{6-7} level. The motor neurons which innervate the elbow flexor muscles, biceps brachii, and brachialis, exit the spinal cord in the C_{5-6} region, whereas the motor neurons for the elbow extensors, triceps brachii, exit in the C_{7-8} region (20). The rowing action of Rowcycle propulsion relies heavily on the elbow flexors. Propulsion of the Handbike utilizes both the elbow flexors and extensors, and therefore, probably overtaxed this subject's proportionally weaker extensor muscles.

CONCLUSION

Regardless of experimental condition, mean within-subject differences in speed, HR, $\dot{V}O_2$, $\dot{V}_E$, or NLEC between the Handbike and Rowcycle were small and not statistically significant. Both vehicles appear to have potential for use in aerobic conditioning programs which are designed for persons restricted to upper body exercise. Locomotive economy, as indicated by NLEC, was significantly related to vehicle preference. It was concluded that locomotive economy may be an important component in determining an individual's preference for one upper body–powered vehicle over another. This relationship should be explored further for possible applications in design and testing of mobility devices.

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REFERENCES

1. Bauman WA. The endocrine system. In: Whiteneck GG, Charlifue SW, Gerhart KA, et al., eds. Aging with spinal cord injury. New York: Demos Publications, 1993.
2. Cowell LL, Squires WG, Raven PB. Benefits of aerobic exercise for the paraplegic: a brief review. Med Sci Sports Exerc 1986:18:501–8.
3. Glaser RM, Davis GM. Wheelchair dependent individuals. In: Franklin BA, Gordon S, Timmis GC, eds. Exercise in modern medicine, Baltimore: Williams & Wilkins, 1989:237–67.
4. Leech B. 1993 Handcycle survey. Sports 'N Spokes 1993:Nov/Dec:26–34.
5. Robbins S. Pedal power revisited. Sports 'N Spokes 1989:Sept/Oct:32–5.
6. Schwandt DF. Para-bike: an arm-powered bicycle. In: Proceedings of the 6th Annual RESNA Conference, San Diego, CA, Washington, DC: RESNA Press, 1983:378–80.
7. Schwandt DF. Handbike training manual. Palo Alto, CA: VA Rehabilitation Research and Development Center, 1987.
8. Schwandt D, Leifer L, Axelson P, Anderson J, Glenn AM. The handbike: an arm-powered bicycle. Rehabilitation Research and Development 1988 Progress Report, Palo Alto, CA: VA Medical Center, 1988 (Report 51):115–6.
9. Crase N. The rowcycle: an evaluation. Sports 'N Spokes 1987:Nov/Dec:65–7.
10. Langbein WE, Robinson CJ, Kynast L, Fehr L. Calibration of a new wheelchair ergometer: the Wheelchair Aerobic Fitness Trainer. IEEE Trans Rehabil Eng 1993:1:49–58.
11. Langbein WE, Fehr L. Research device to preproduction prototype: a chronology. J Rehabil Res Dev 1994:30:436–42.
12. Langbein WE, Maki KC, Edwards LC, Hwang MH, Sibley P, Fehr L. Initial clinical evaluation of a wheelchair ergometer for diagnostic exercise testing. J Rehabil Res Dev 1994:31:317–25.
13. The Preventive and Rehabilitative Exercise Committee of the American College of Sports Medicine. Principles of exercise prescription. In: Guidelines for graded exercise testing and exercise prescription. 4th ed. Philadelphia: Lea & Febiger, 1991.
14. Wasserman K, McIlroy MB. Comparison of continuous and discontinuous protocols for the determination of peak oxygen uptake in arm cranking. Eur J Appl Physiol 1964:51:3–6.
15. McArdle WD, Katch FI, Katch VL. Exercise physiology: energy, nutrition and human performance. Philadelphia: Lea & Febiger, 1986:635–49.
16. Glaser RM, Sawka MN, Wilde SW, Woodrow BK, Suryaprasad AG. Energy cost and cardiopulmonary responses to wheelchair locomotion and walking on tile and on carpet. Paraplegia 1981:19:220–6.
17. Kirkwood BR. Essentials of medical statistics. London: Blackwell Scientific Publications, 1988.
18. Medrano D. Energy cost of row-cycling for paraplegics (Thesis). Fresno, CA: California State University, 1988.
19. Hooker SP, Wells C. Effects of low- and moderate-intensity training in spinal cord-injured persons. Med Sci Sports Exerc 1989:1:18–22.
20. Donovan WH, Bedbrook G. Comprehensive management of spinal cord injury. Clin Symp 1982:34:1–36.

ABSTRACTS OF RECENT LITERATURE

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Abstracts are drawn primarily from the orthotics and prosthetics literature. Selections of articles were made from these journals:

Archives of Physical Medicine and Rehabilitation
Journal of Prosthetics and Orthotics
Medizinisch-Orthopädische Technik
Physical Therapy
Prosthetics and Orthotics International

PROSTHETICS, ORTHOTICS, AND RELATED TOPICS

Benchmark Data for Elderly, Vascular Trans-Tibial Amputees After Rehabilitation. Hubbard WA, McElroy GK. Reprinted from *Prosthet Orthot Int* 18:142–149, 1994.

Benchmark data for lower limb amputees is often limited to young subjects who have had their amputations as the result of trauma. The majority of trans-tibial amputees rehabilitated are, however, elderly vascular amputees who may have different gait characteristics than their younger counterparts. Without biomechanical analyses to provide such benchmark data for this group it is not possible to compare the effects of different rehabilitation programmes, gait training regimens, or prosthetic devices.

Twenty elderly vascular trans-tibial amputees rehabilitated at The Queen Elizabeth Centre, Ballarat, Australia and at least six months post-amputation were measured in respect of kinetic and kinematic parameters, and relationships between gait speed, consistency, and function were demonstrated. Further, an unexplained vertical ground reaction force pattern was demonstrated in faster, more functional amputees.

Bioenergetic Comparison of a New Energy-Storing Foot and SACH Foot in Traumatic Below-Knee Vascular Amputations. Casillas J-M, Dulieu V, Cohen M, et al. Reprinted from *Arch Phys Med Rehabil* 76:39–44, 1995 (©1995 by the American Congress on Rehabilitation Medicine and the American Academy of Physical Medicine and Rehabilitation).

In this study, the metabolic performances of a new energy-storing foot (Proteor) and of the solid-ankle cushion heel (SACH) are compared. Twelve patients with traumatic below-knee amputations (mean age: 50.0 ± 19.9 years) and 12 patients with vascular below-knee amputations (mean age: 73 ± 7 years) were studied. Oxygen uptake ($\dot{V}O_2$) was measured in all the subjects on a walkway at a self-selected velocity; only the subjects with traumatic amputation were tested on a level treadmill (progressive speed: 2.4–4 and 6 km/h), and then in two randomized trials: incline (+5%) and decline walking treadmill test at 4 km/h. Vascular explorations were done in the vascular patients: distal pressure measurements, pulse plethysmography, transcutaneous oxygen tension. Free walking was improved in subjects with traumatic amputation using the energy-storing foot (+6%), with a better bioenergetic efficiency (0.24 ± 0.4mL/kg·m vs 0.22 ± 0.04mL/kg·m). However, in subjects with vascular amputation, this foot did not produce an increased free velocity nor an improved energy cost. During the level treadmill test, the traumatic amputee subjects showed a decrease of energy expenditure with the new prosthetic foot, more significant at sufficient speed (4km/h): 17.00 ± 3.42 vs 14.67 ± 2.05 mL/kg/min ($p < .05$). The same effect is shown during the incline (19.31 ± 2.80 vs 16.79 ± 2.32mL/kg/min—$p < .02$) and decline walking tests (14.13 ± 3.64 vs 11.81 ± 1.54mL/kg/min—$p < .02$). There is no significant difference in cardiocirculatory effects between the two types of prosthetic foot. Despite a lower velocity, the subjects with vascular amputation exceed 70% of the maximal heart rate, with the cardiocirculatory factor being the main cause of walking restriction. The energy-storing foot should be reserved for active and fast walkers, whereas the SACH foot seems more suitable for elderly patients with amputation with a slow walk.

Biomechanical Assessment of Quiet Standing and Changes Associated with Aging. Panzer VP, Bandinelli S, Hallett M. Reprinted from *Arch Phys Med Rehabil* 76:151–157, 1995 (©1995 by the American Congress on Rehabilitation Medicine and the American Academy of Physical Medicine and Rehabilitation).

The kinematics of standing balance were analyzed in 24 normal subjects, aged 21 to 78 years, to examine differences attributable to age, visual input, and sex. Movements of individual body segments, displacement of the center of gravity (COG), and position of the center of pressure (COP) were measured, and total path length and variability about the mean position were derived from the resulting values. Aging was associated with an increase in variability of the COG, head, and hip, but not in path length. The changes, which may be clinically interpreted as excess postural sway, do not show stability deficits as a consequence of aging. On the contrary, older subjects seem to adopt a postural control strategy that achieves comparable stability during quiet standing. Eye closure increased the anterior-posterior COP path length without corresponding changes in the COG, indicating an increase in small accelerations without associated instability. There was more medial-lateral movement in women than in men. Quantitative electromyographic measures showed that, in general, quiet standing requires very little muscular activity. We conclude that the task of quiet standing produces no evidence of postural instability concurrent with aging. The altered postural control strategy may be less effective when balance is suddenly or severely compromised.

Biomechanisch-ganganalytische Bewertung von Prothesenfussen (The Assessment of Prosthetic Feet with Reference to Biomechanics and Gait Analysis). Blumentritt S, Scherer HW, Wellershaus U. Reprinted from *Med Orth Tech* 114:287–292, 1994.

The number of prosthetic feet available for amputees has increased considerably in the recent past. Knowledge of the relevant functional characteristics of prosthetic feet, especially those effecting gait, simplifies the choice of a foot which would suit the patient's needs best. The effect of various foot types on gait has been determined on three below-knee amputees by means of gait analysis (KISTLER force plate, PRIMAS system). Depending on the aims of rehabilitation envisaged, suitable feet for below- and above-knee amputees are available for individual requirements.

Critical Evaluation of Clinical Research. Katz RT, Campagnolo DI, Goldberg G, et al. Reprinted from *Arch Phys Med Rehabil* 76:82–93, 1995 (©1995 by the American Congress on Rehabilitation Medicine and the American Academy of Physical Medicine and Rehabilitation).

This focused review contains a suggested core of material that will help residents or practicing physiatrists critically review research papers published in the medical literature. Before accepting the results of a clinical trial, physiatrists must critique the experimental methods and study design carefully to decide whether to include these new ideas into their clinical practice. Medical research relies on statistical methodology, and statistics pervade the medical literature. This article begins with an introduction to rudimentary statistics. Fortunately, most studies depend on a rather small body of statistical concepts. The elements of experimental design—clinical trials, randomization, single-subject design, meta-analysis, epidemiological studies—are presented in a concise review. Finally, the elements of statistics and experimental design are integrated into a step-by-step method strategy for reading the medical literature.

Einsatz von Hydraulisch und Pneumatisch Gesteuerten Prothesen-Kniegelenken (Application of Hydraulically and Pneumatically Controlled Prosthetic Knee Joints). Milde L, Scherer HW. Reprinted from *Med Orth Tech* 114:292–299, 1994.

In fitting above-knee amputees, the choice of the most suitable prosthetic knee joint is a very important matter. Technical means of locking the joint during the stance phase very from shifting the knee axis posteriorly through poly-centric systems to hydraulic mechanisms. Especially in the case of active patients, hydraulic and pneumatic designs are recommended.

Age, general condition and the condition of the stump mainly influence the efficiency, and these factors must be taken into consideration when prescribing a prothesis.

Four-bar Linkage Prosthetic Knee Mechanisms: Kinematics, Alignment and Prescription Criteria. Radcliffe CW. Reprinted from *Prosthet Orthot Int* 18:159–173, 1994.

Four-bar linkage knee mechanisms for the trans-femoral amputee are widely available, but although they may offer functional advantages to certain amputees, they are fitted in a limited number of cases. It may be assumed that one reason for this is that persons responsible for prescription and fitting may not be familiar with the kinematic charac-

teristics and possible advantages of such mechanisms and are reluctant to use a device which they do not understand completely.

This paper will describe the kinematics of several types of four-bar mechanisms, and discuss the differences and prescription criteria for three different classes of four-bar linkage mechanisms currently available for fitting to amputees. Before beginning the discussion of four-bar prosthetic knees, it will be helpful to review some fundamental concepts.

Four Methods for Characterizing Disability in the Formation of Function Related Groups. Stineman MG, Hamilton BB, Granger CV, et al. Reprinted from *Arch Phys Med Rehabil* 75:1277–1283, 1994 (©1994 by the American Congress on Rehabilitation Medicine and the American Academy of Physical Medicine and Rehabilitation).

The Functional Independence Measure-Function Related Groups (FIM-FRGs) were developed to classify medical rehabilitation inpatients into homogeneous groups based on length of stay (LOS). Patients are first grouped into clinically relevant rehabilitation impairment categories, then by functional status, as expressed by the FIM, and in certain cases by patient age. The statistical approach used to form the final groupings was a recursive partitioning algorithm applied to the FIM scores and patient age within impairment category. This analysis compares four FIM-FRG classification schemes developed from four scale sets that combine FIM items differently: (1) use of the 18 FIM items as separate variables, (2) the combination of FIM items into six clinical subscales, (3) the combination of the six clinical subscales into motor and cognitive subscales, and (4) the combination of all FIM items into a single scale. The FIM-FRG schemes explain similar amounts of variance in the logarithm of LOS and contain approximately equal numbers of FRGs. The motor and cognitive subscale scheme is recommended for use in payment, however, this scheme and the other schemes have additional uses. Each FRG scheme provides different insight into the clinical relationship between disability and LOS.

Gehen-Laufen mit Prothesen (Walking-Running with Prostheses. Kristen H, Kastner J. Reprinted from *Med Orth Tech* 114–279–282, 1994.

Differences in the use of the terms "walking" and "running" are pointed out. The functional changes in these two forms of locomotion are described. Compensation possibilities are shown, and new technical developments to substitute the required function are discussed.

Incidence of Skin Breakdown and Higher Amputation After Transmetatarsal Amputation: Implications for Rehabilitation. Mueller MJ, Allen BT, Sinacore DR. Reprinted from *Arch Phys Med Rehabil* 76:50–54, 1995 (©1995 by the American Congress on Rehabilitation Medicine and the American Academy of Physical Medicine and Rehabilitation).

The purpose of this study of patients with transmetatarsal amputation (TMA) is to describe multiple patient characteristics, including the incidence of subsequent skin breakdown and higher amputation, that may influence rehabilitation treatment and outcomes. Data were gathered on all patients having a TMA at this facility between April 1989 and September 1993. One hundred twenty TMAs were performed on 107 patients with a mean age of 62.4 ± 13.8 years. There were 55 men and 52 women. Thirteen patients (12%) had a bilateral TMA. Twenty-nine patients (27%) developed skin breakdown. Of these, 48% occurred within the first 3 months after surgery. Thirty patients (28%) required a higher amputation. Of these, 60% occurred in the first month after TMA. In addition, this group of patients had a high incidence of diabetes mellitus (77%), hypertension (54%), electrocardiogram (EKG) abnormalities (60%), congestive heart failure (22%), and prior ipsilateral vascular surgery (51%). These results indicate that patients with TMA often present with a complicated medical condition and that they are at high risk of skin breakdown or higher amputation, especially in the first 3 months after surgery. The investigators conclude that patients with TMA may benefit from a rehabilitation program emphasizing protection of the residuum during their return to functional activities. Additional research is needed to determine optimal acute and long-term rehabilitation of patients with TMA.

Increased Muscular Strength in Paralyzed Patients After Spinal Cord Injury: Effect of Beta-2 Adrenergic Agonist. Signorile JF, Banovac K, Gomez M, et al. Reprinted from *Arch Phys Med Rehabil* 76:55–58, 1995

The administration of β-2 adrenergic agonists in experimental animals result in an increased strength of skeletal muscle. In this study, we evaluated whether a β-2 adrenergic agonist, metaproterenol, had an effect on muscle size and strength in a group of patients with muscular atrophy following spinal cord injury. Ten male subjects were randomly divided into 2 groups and agreed to participate in a prospective, double-blind, placebo-controlled, and crossover study. Metaproterenol (80 mg/day), or placebo, was administered orally for a period of 4 weeks. Muscle strength was measured by a force transducer interfaced with a microcomputer. Muscle size was calculated and expressed as a cross-sectional area of upper arm and forearm using a formula. Metaproterenol induced a significant increase of muscle strength in both groups of subjects, compared with placebo ($p < .001$). Similarly, there was an increase in a muscle size in the forearm following the administration of metaproterenol. Our data indicate that β-2 adrenergic agonists may improve both muscle strength and size in patients with muscular atrophy following spinal cord paralysis.

The Influence of Body Size on Linear Measurements Used to Reflect Cervical Range of Motion. Chibnall JT, Duckro PN, Baumer K. Reprinted from *Phys Ther* 74:1134–1137, 1994.

Background and Purpose. The purposes of the study were to demonstrate that linear measurements of cervical range of motion are confounded by differences in body size when comparisons are made between groups and to introduce an alternative measure of range of motion that adjusts for variation in body size. Subjects. The sample consisted of 42 subjects (25 female, 17 male) with chronic posttraumatic headaches. Methods. Using a tape measure, a physical therapist measured the distance between two anatomical landmarks with the subject's neck in the anatomical neutral position and with the subject's neck fully laterally flexed, rotated, and forward flexed. Range of motion was effected with two measurements: the distance between the landmarks at full flexion/rotation and the proportion of distance traversed between the landmarks. Results. The end-of-range values were significantly correlated with body size. No significant correlations emerged between the proportion-of-distance values and body size. Conclusion and Discussion. Linear measurements of cervical motion are potentially confounded by body size when subjects of nonequivalent size are compared. Proportion-of-distance measurement is presented as a more valid alternative to end-of-range measurement.

Inter-Rater Agreement and Stability of Functional Assessment in the Community-Based Elderly. Ottenbacher KJ, Mann WC, Granger CV, et al. Reprinted from *Arch Phys Med Rehabil* 75:1297–1301, 1994 (©1994 by the American Congress on Rehabilitation Medicine and the American Academy of Physical Medicine and Rehabilitation).

The purpose was to examine the inter-rater agreement and test-retest stability of the Functional Independence Measure (FIM) and the Instrumental Activities of Daily Living Scale (IADL) from the Multidimensional Functional Assessment of Older Adults. These two instruments were administered to 20 older persons living in the community. Two experienced raters administered the assessment instruments over either a short (7–10 days) or long (4 to 6 week) interval. The intraclass correlation (ICC) was used to analyze the data. ICC values were computed for agreement between and within raters and across short and long intervals. ICC values for interrater agreement and stability ranged from 0.90 to 0.99. The relation between scores on the FIM and IADL scale was also examined. The analysis produced an r value of 0.85, suggesting a positive statistical relationship among the items assessed. The high ICC values indicate that the Functional Independence Measure Instrument and IADL scale of the Multidimensional Functional Assessment of Older Adults provide consistent information across two experienced raters and over time when used with a sample of elderly persons residing in the community.

Kinematics of Wheelchair Propulsion in Adults and Children with Spinal Cord Injury. Bednarczyk JH, Sanderson DJ. Reprinted from *Arch Phys Med Rehabil* 75:1327–1334, 1994 (©1994 by the American Congress on Rehabilitation Medicine and the American Academy of Physical Medicine and Rehabilitation).

This study examined the kinematic features of wheelchair propulsion in two neurologically matched groups of adults and children with uncomplicated spinal cord injury. The average mass and age of the pediatric group was much smaller than the adult group (37.4kg and 11.3 years vs 68.5kg and 33.5 years). Each subject propelled his/her own

chairs and new, low-mass wheelchairs at a steady, nominal speed of 2 m/sec across a level surface. Three dimensional video analysis determined the movement of upper body angles (elbow, shoulder, trunk, and shoulder abduction) based on reflective markers placed on the subjects' shoulder, elbow, wrist, and hip joints. Analysis of the temporal factors showed that although the average group overground velocities of the adult group (2.4m/sec) were significantly greater than the pediatric group (2.3m/sec), the two groups spent comparable proportions of the wheeling cycle in propulsion (24%). Analysis of the angular kinematics (elbow, shoulder, and shoulder abduction angular changes over a time normalized wheeling cycle) showed that whereas the pediatric group did show significant absolute angular differences from the adult group, the angular changes over time were the same in both groups. The implications of this work are that, for the first time, it can be said that children propel their wheelchairs in the same manner as adults. In addition, these data were similar to those previously reported in athletic adult populations. We conclude that published data from adult wheelchair users may be applied to pediatric wheelchair users, thus providing a basis for pediatric wheelchair prescription. Further studies using kinetic measurements over longer time periods may be necessary to elucidate other potential similarities between the adult and pediatric wheeling style.

Kosmetik von Beinprothesen-Kosmetische Beinprothesen (Cosmesis in Lower Extremity Prosthetics-Aesthetic Lower Extremity Prostheses. Baumgartner R, Botta P, Turk K. Reprinted from *Med Orth Tech* 114:267–271, 1994.

The common aim of amputation surgery and prosthetic technology lies in restoring not only the function but also the outward appearance, the so-called cosmesis. Surgical measures should prevent over-lengths, over-widths and misalignment of axes.

By virtue of modular components, modern fittings, materials and manufacturing methods, prosthetic technology can now produce artificial legs of high cosmetic quality. This includes such devices made solely as camouflage, for temporary or permanent use.

Especially in modular prostheses, function and cosmesis are not necessarily in union. Generally, cosmesis is a hinderance to function. It is therefore demanded to incorporate both requirements simultaneously in a prosthesis.

Microprozessorgesteuerte Schwungphasensteuerung fur Oberschenkelprothesen (Microprocessor Controlled Swing Phase for Above Knee Prostheses. Fitzlaff G. Reprinted from *Med Orth Tech* 114:299–303, 1994.

Swing phase control introduces a damping resistance that prevents exaggerated lifting of the heel and excessive swing of leg in flexion and extension. On change of walking speed, conventional swing phase controls require more force and energy. In microprocessor controlled swing phase, the damping resistance is adjusted during extension by a set valve orifice. The damping resistance during knee flexion, however, is adjusted by means of a linear step motor according to the momentary walking speed. In this manner, natural and energetically efficient walking with the prosthesis is enabled.

Optimierung von Prothesenschaften mit dem Air Contact System (ACS) (Optimization of Prosthetic Sockets with the Air Contact System (ACS). Pohlig K. Reprinted from *Med Orth Tech* 114:272–276, 1994.

The volume of an above-knee stump is not constant. Through illness, weight loss, growth, or sport activities, it no longer matches the volume of the socket, the position adopted by the stump depends on its length–the higher the amputation level, the more abduction and external rotation issues. The Air Contact System gives the amputee a prosthetic socket that adjusts itself according to the requirements of a given situation. On filling the air chambers it promotes an internal rotation and adduction of the femur. The cross-section of the socket can be varied. The longitudinal load bearing at the end of the stump can be chosen to meet individual needs and can also be made stronger than was possible up to now.

Orientierende Untersuchung uber Energiespeicherung und Energierunckgabe von Prothesenfussen (Pilot Investigations on Prosthetic Feet Employing Energy Storage Systems). Saccetti R, Schmidl H, von Groningen M. Reprinted from *Med Orth Tech* 114:293–295, 1994.

Six different prosthetic feet were tested on a single person. Gait analysis permitted the energy input and output to be determined during the walking phase. Because the best results obtained did not correspond with the subjective description given by the amputee, it is of major importance to take the patient's preferences into account when choosing a foot.

Outcomes of Post-Spinal Cord Injury Marriages. DeVivo MJ, Hawkins LN, Richards JS, et al. Reprinted from *Arch Phys Med Rehabil* 76:130–138, 1995 (©1995 by the American Congress on Rehabilitation Medicine and the American Academy of Physical Medicine and Rehabilitation).

The purpose of this study was to compare the divorce rate among persons who got married after spinal cord injury with that of the non-spinal cord-injured population of comparable age and gender and to identify factors associated with increased likelihood of divorce. The study included 622 persons enrolled in the National Spinal Cord Injury Statistical Center data set since 1973. These persons were followed between 1 and 15 years after their marriage (mean = 3.5 years). The status of each marriage was determined at the time of their most recent routine annual follow-up examination. Overall, 126 divorces occurred, whereas 74 were expected, based on 2,190 person-years of follow-up and age-sex-specific annual divorce rates for the United States population. Men and remarried persons had divorce rates 2.07 times and 1.80 times higher, respectively, than women and persons married for the first time. The divorce rate was 1.85 times higher among persons without college educations and was lower for persons with lumbosacral injuries than for persons with higher injury levels. In general, the impact of spinal cord injury appears to be almost as great on postinjury marriages as it is on preexisting marriages. However, this study yields descriptive rather than causal information. Other factors must be identified before a clinically useful model to predict persons at high risk for divorce can be developed.

Prosthetic Profile of the Amputee Questionnaire: Validity and Reliability. Gauthier-Gagnon C, Grise M-C. Reprinted from *Arch Phys Med Rehabil* 75:1309–1314, 1994 (©1994 by the American Congress on Rehabilitation Medicine and the American Academy of Physical Medicine and Rehabilitation).

The Prosthetic Profile of the Amputee (PPA) is a clinical follow-up questionnaire that measures the factors potentially related to prosthetic use and the actual use of the prosthesis by people with a lower extremity amputation. This profile is a report on the psychometric properties of the questionnaire. The PPA was administered, on two occasions, to 89 adults with lower extremity amputations. Test-retest data analysis demonstrated that the questionnaire was reliable in terms of repeatability. Strong test-retest agreements were obtained. When the PPA was against the Reintegration to Normal Living (RNL) index, results supported the presence of construct validity. Data confirmed convergence of analogous constructs of the PPA questionnaire and the RNL index and showed evidence of discrimination between the more distantly related constructs. Based on the results of this study, we conclude that the PPA questionnaire is reliable and valid for clinical and research use.

Recovery of Walking Function in Stroke Patients: The Copenhagen Stroke Study. Jorgensen HS, Nakayama H, Raaschou HO, et al. Reprinted from *Arch Phys Med Rehabil* 76:27–32, 1995 (©1995 by the American Congress on Rehabilitation Medicine and the American Academy of Physical Medicine and Rehabilitation).

Time course and degree of the recovery of walking function after stroke and the influence of initial lower extremity (LE) paresis were studied prospectively in a community-based population of 804 consecutive acute stroke patients. Walking function and degree of LE paresis were assessed weekly using the Barthel index and the Scandinavian Neurological Stroke scale, respectively. Initially, 51% had no walking function, 12% could walk with assistance, and 37% had independent walking function. At the end of rehabilitation, 21% had died, 18% had no walking function, 11% could walk with assistance, and 50% had independent walking function. Recovery of walking function occurs in 95% of the patients within the first 11 weeks after stroke. The time and the degree of recovery are related to both the degree of initial impairment of walking function and to the severity of LE paresis, $p < .0001$. A valid prognosis of walking function in patients with initially *no/mild/moderate* leg paresis can be made in 3 weeks, and further recovery should not be expected after 9 weeks. A valid prognosis of walking function in patients with initially severe leg paresis or paralysis can be made in 6 weeks, and further improvement of walking function should not be expected later than 11 weeks after stroke.

Rehabits: A Common Language of Functional Assessment. Fisher WP, Harvey RF, Taylor P, et al. Reprinted from *Arch Phys Med Rehabil* 76:113–122, 1995 (©1995 by the American Congress on Rehabilitation Medicine and the American Academy of Physical Medicine and Rehabilitation).

Probabilistic measurement models offered by Rasch and others can be used to link different functional assessment instruments into a single measurement system. This study assessed 54 subjects (diagnoses: 8 brain injuries, 7 neuromuscular, 22 musculoskeletal, 7 spinal cord, 10 stroke) admitted to a free-standing rehabilitation hospital at admission and discharge using both the Functional Independence Measure (FIM) and the Patient Evaluation and Conference System (PECS). Thirteen FIM and 22 PECS motor skills items were scaled together into a 35-item instrument, providing scale values for all items in the same unit of measurement. Separate FIM and PECS measures produced for each subject correlate .94 and .91 ($p < .0001$), respectively, with the cocalibration measures, and 0.91 ($p < .0001$) with each other. Either instrument's ratings are easily and quickly converted into the other's using the common unit of measurement, the rehabit (rehabilitation measuring unit). This article argues that the stability of the PECS and FIM item difficulty estimates over thousands of subjects, dozens of hospitals, hundreds of raters, and years of assessment is convincing evidence in support of the widespread use of their cocalibrated, common scale values as a functionometric ruler.

Rheumatoid Arthritis: New Approaches for Its Evaluation and Management. Semble EL. Reprinted from *Arch Phys Med Rehabil* 76:190–201, 1995 (©1995 by the American Congress on Rehabilitation Medicine and the American Academy of Physical Medicine and Rehabilitation).

Rheumatoid arthritis is a chronic, progressive disease with a long-term outcome characterized by significant morbidity, loss of functional capacity, and increased mortality. The cornerstone of therapy includes the appropriate melding of pharmacological, rehabilitative, and surgical treatments. New developments in the care of patients with rheumatoid arthritis have focused on aggressive pharmacological therapy early in the course of the illness, ongoing assessment of disease activity and patient function, and a better understanding of the role of rehabilitative techniques such as therapeutic exercise and behavioral approaches to education. This article synthesizes information from studies on recent advances in the management of rheumatoid arthritis outlining diagnosis and assessment, disability issues, outcome studies, current status of traditional and experimental pharmacological therapies, and new strategies of nonpharmacological treatments aimed at the clinician challenged by this fascinating disorder.

Silicone-Only Suspension (SOS) with Socket-Loc and the Ring for the Lower Limb. Haberman LJ. Reprinted from *J Prosthet Orthot* 7:2–14, 1995.

This article discusses the previous and current uses of silicone liners with the Silicone-Only Suspension (SOS) technique. This technique has been enhanced by the use of injection-molded silicone liners, Socket-Loc (patent pending), and the Ring. Previously reported problems with SOS included unwanted internal socket rotation and intermittent loss of suction suspension. Socket-Loc and the Ring have eliminated these difficulties while retaining the many advantages of the SOS technique.

Standing Balance in Trans-Tibial Amputees Following Vascular Disease or Trauma: A Comparative Study with Healthy Subjects. Hermodsson Y, Ekdahl C, Persson BM, et al. Reprinted from *Prosthet Orthot Int* 18:150–158, 1994.

Standing balance measured as sway and standing time both on one and two legs, was studied by use of a stable force platform (Kistler) in 36 patients aged 48–87 years with trans-tibial amputation and 27 healthy subjects matched for age. The aim of the study was to compare postural function in standing in two groups with unilateral trans-tibial amputations, separating vascular disease from trauma. Results revealed that the vascular group had a significantly increased sway in the lateral direction compared with the healthy group, when standing on both feet close together for 30 seconds, looking straight ahead or blindfolded (p values ranging from 0.003 to 0.02). In the sagittal direction the trauma amputees had a significantly decreased sway when looking straight ahead, compared to the vascular and healthy groups (p values = 0.03). No significant differences in the lateral or sagittal direction were seen among the three groups when comparing standing on one leg. There was a significant difference, however, in the standing time in the one-leg standing test of the vascular group when compared with the trauma and healthy groups (p values ranging from 0.0009 to 0.02). In contrast to the vascular group, all subjects in the trauma and healthy groups from 48 to 59 years could stand on the healthy leg for 30 seconds when looking straight ahead, and from 60 to 79 years they could stand for 5 seconds. None in the vascular or trauma group older than 80 years could stand on the healthy leg for 5 seconds. The standing balance of the vascular amputees was found to be inferior to that of the trauma amputees. In conclusion, vascular and trauma trans-tibial amputees should not be considered as an entity in test situations or rehabilitation programmes.

The Use of Splints in the Treatment of Joint Stiffness: Biologic Rationale and an Algorithm for Making Clinical Decisions. McClure PW, Blackburn LG, Dusold C. Reprinted from *Phys Ther* 74:1101–1107, 1994.

The purposes of this article are (1) to discuss the rationale for using splints to increase range of motion (ROM) and (2) to describe an algorithm that can guide therapists' clinical decisions when splints are used to treat patients who have limited ROM. The primary rationale for using splints is to apply relatively long periods of tensile stress to shortened connective tissues to induce tissue lengthening through biologic remodeling. The process of remodeling is contrasted with more temporary mechanical phenomena that occur in biologic tissues. The proposed algorithm guides the use of splints based on measurements of pain and ROM. We describe three variables of splint use that may be adjusted: frequency, duration, and intensity. The relative importance of each of these variables is discussed. The algorithm is not joint or injury specific and requires continual modification of splint use based on a patient's response to treatment. Deciding which patients are appropriate for end-range splinting and deciding when to discontinue splint use are also discussed.

Why Rehabilitation Research Does not Work (As Well as We Think It Should). Ottenbacher KJ. Reprinted from *Arch Phys Med Rehabil* 76:123–129, 1995 (©1995 by the American Congress on Rehabilitation Medicine and the American Academy of Physical Medicine and Rehabilitation).

Establishing treatment effectiveness is a high priority for rehabilitation research. The use of traditional quantitative null hypotheses to achieve this priority is reviewed. Three problems are identified in the analysis and interpretation of investigations based on statistical testing of hypotheses: (1) confusion of clinical and statistical significance, (2) low statistical power in detecting clinically important results, and (3) a failure to understand the importance of replication in developing a knowledge base for rehabilitation practice. Technical aspects associated with each problem are reviewed and examples presented illustrating the impact of low statistical power and the results of misinterpreting statistical significance tests. Several specific recommendations are made to improve the clinical usefulness of quantitative research conducted in rehabilitation.

BOOK REVIEWS

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Functional Electrical Stimulation for Ambulation by Paraplegics: Twelve Years of Clinical Observations and System Studies. Daniel Graupe, PhD and Kate H. Kohn, MD. Malabar, FL: Krieger Publishing Company, 1994. 194 pp.

This book was written by two investigators who have pioneered and led the development of the "Parastep" functional electrical stimulation (FES) ambulation system for use by persons with spinal cord injury (SCI), the first such system to become commercially available in the United States, and to be approved by the Food and Drug Administration (FDA). Following introductory chapters describing the background of their work and the physiology of nerve excitation, the book focuses on the authors' 12-year experience in the field of FES. In the first half of the book, the numerous steps in the development of computerized FES systems for ambulation are detailed, including the engineering and design of system parameters, operational components, and safety features. The second half of the book is more clinically oriented and is based on the authors' and their coinvestigators' experience with 100 persons with SCI who have been trained to use the FES system for short distance ambulation. This part of the book is particularly useful for clinicians who evaluate, select, and train persons to use the "Parastep" ambulation system. The clinical results and the authors' own observations of 20 subjects using the system at the Michael Reese Hospital in Chicago are addressed in detail and the results of 80 other subjects who participated in a multisite clinical trial at 12 different health institutions are described.

This book is a fascinating description of a remarkable accomplishment by determined investigators who developed, without fanfare, a simple but clinically workable FES system for ambulation. This book is, therefore, a must for all people who are involved in FES research or clinical application of this technology. It contains information that is not readily available elsewhere, and provides a valuable inspiration to other developers of devices and technology for persons with disabilities. Its primary weaknesses lie in inadequate editing, unreferenced statements and limited discussion of the work of others in the field of FES.

PUBLICATIONS OF INTEREST

This list of references offers *Journal* readers significant information on the availability of recent rehabilitation literature in various scientific, engineering, and clinical fields. The *Journal* provides this service in an effort to fill the need for a comprehensive and interdisciplinary indexing source for rehabilitation literature.

All entries are numbered so that multidisciplinary publications may be cross-referenced. They are indicated as *See also* at the end of the categories where applicable. A listing of the periodicals reviewed follows the references. In addition to the periodicals covered regularly, other publications will be included when determined to be of special interest to the rehabilitation community. To obtain reprints of a particular article or report, direct your request to the appropriate contact source listed in each citation.

AMPUTATIONS

1. Benchmark Data for Elderly, Vascular Trans-Tibial Amputees After Rehabilitation. Hubbard WA, McElroy GK, *Prosthet Orthot Int* 18(3):142–149, 1994.
Contact: Wendy A. Hubbard, Chief Physiotherapist, The Queen Elizabeth Centre, Ballarat, P.O. Box 199, Ballarat, Victoria, Australia 3350

2. Diabetic Sourcebook. Bellenir K, Dresser P, eds., *Detroit, MI: Omnigraphics* 1995:827 pages.
Contact: Karen Bellenir, c/o Omnigraphics, Penobscot Building, Detroit, MI 48226

3. Gait of Unilateral Transfemoral Amputees. Boonstra AM, et al., *Scand J Rehabil Med* 26(4):217–223, 1994.
Contact: A.M. Boonstra, MD, Dept. of Rehabilitation Medicine, University Hospital Groningen, P.O. Box 30.001, NL-9700 RB Groningen, The Netherlands

4. Incidence of Skin Breakdown and Higher Amputation After Transmetatarsal Amputation: Implications for Rehabilitation. Mueller MJ, Allen BT, Sinacore DR, *Arch Phys Med Rehabil* 76(1):50–54, 1995.
Contact: Michael J. Mueller, PhD, PT, Assist. Prof., Washington University School of Medicine, Campus Box 8502, 660 S. Euclid Ave., St. Louis, MO 63110–1093

5. Lower Extremity Amputation in a Marine Veteran 22 Years Following Vascular Injury in Vietnam. Coy J, *Milit Med* 159(11):718–719, 1994.
Contact: Col. Jim Coy, MC, FS, USUHS, Clinical Asst. Prof., Military/Emergency Medicine (IMA) and Medical Consultant USASOC, Harry S. Truman Veterans Hospital, 800 Hospital Dr., Columbia, MO 65201–5297

6. Patterns of Recovery for Lower Limb Amputation. Ham R, de Trafford J, Van de Ven C, *Clin Rehabil* 8(4):320–328, 1994.

Contact: Rosalind Ham, Newham Wheelchair Service, St Andrew's Hospital, London E3 3NT, UK

7. Persian Gulf War Amputees: Injuries and Rehabilitative Needs. Dillingham TR, Braverman SE, Belandres PV, *Milit Med* 159(10):635–639, 1994.
Contact: Timothy R. Dillingham, MD, Dept. of Rehabilitation Medicine, The Johns Hopkins University, Baltimore, MD 21213

8. Prosthetic History, Prosthetic Charges, and Functional Outcome of the Isolated, Traumatic Below-Knee Amputee. Smith DG, et al., *J Trauma* 38(1):44–47, 1994.
Contact: Douglas G. Smith, MD, Asst. Prof. of Orthopaedic Surgery, Harborview Medical Center, 325 Ninth Ave., ZA-48, Seattle, WA 98104

ASSISTIVE DEVICES

9. Measuring Quality and Performance of Assistive Technology: Results of a Prospective Monitoring Program. Kohn JG, LeBlanc M, Mortola P, *Assist Technol* 6(2):120–125, 1994.
Contact: Jean G. Kohn, MD, MPH, Rehabilitation Engineering Center, Lucile Salter Packard Children's Hospital at Stanford, 725 Welch Rd., Palo Alto, CA 94304

10. Need for Information on Assistive Devices by Older Persons. Mann WC, et al., *Assist Technol* 6(2):134–139, 1994
Contact: William C. Mann, OTR, PhD, Rehabilitation Engineering Research Center on Aging, SUNY-Buffalo, 515 Kimball Tower, Buffalo, NY 14214

BIOENGINEERING

11. Detection of Brainstem Auditory Evoked Potential by Adaptive Filtering. Chan FHY, et al., *Med Biol Eng Comput* 33(1):69–75, 1995.
Contact: F.H.Y. Chan, Dept. of Electrical & Electronic Engineering, University of Hong Kong, Pokfulam Rd., Hong Kong

12. Four-Bar Linkage Prosthetic Knee Mechanisms: Kinematics, Alignment and Prescription Criteria. Radcliffe CW, *Prosthet Orthot Int* 18(3):159–173, 1994.
Contact: Prof. C.W. Radcliffe, Hosmer Dorrance Corporation, 561 Division St., Campbell, CA 95008-6905

13. Galvanic Vestibular Stimulation for Analysis of Postural Adaptation and Stability. Johansson R, Magnusson M, Fransson PA, *IEEE Trans Biomed Eng* 42(3):282–291, 1995.
Contact: Rolf Johansson, Dept. of Oto-Rhino-Laryngology, Lund University Hospital, S-221 85 Lund, Sweden

14. In-Shoe Foot Pressure Measurement System Utilising Piezoelectric Film Transducers. Nevill AJ, Pepper MG, Whiting M, *Med Biol Eng Comput* 33(1):76–81, 1995.
Contact: Dr. M.G. Pepper, Electronic Engineering Laboratory, University of Kent at Canterbury, Kent CT2 7NT, UK

BIOMECHANICS

15. Bone Strain Gage Data and Theoretical Models of Functional Adaptation. Mikic B, Carter DR, *J Biomech* 28(4):465–469, 1995.

Contact: Borjana Mikic, Rehabilitation Research and Development Center, Palo Alto VA Medical Center (153), 3801 Miranda Ave., Palo Alto, CA 94304

16. Design Criteria for the Reduction of Shear Forces in Beds and Seats. Goossens RHM, Snijders CJ, *J Biomech* 28(2):225–230, 1995.

Contact: Richard H.M. Goossens, Erasmus University Rotterdam, Faculty of Medicine, Dept. of Biomedical Physics and Technology, P.O. Box 1738, 3000 DR Rotterdam, The Netherlands

17. Impression Technique for the Assessment of Oedema: Comparison with a New Tactile Sensor that Measures Physical Properties of Tissue. Lindahl OA, Omata S, *Med Biol Eng Comput* 33(1):27–32, 1995.

Contact: Olof A. Lindahl, Dept. of Surgery and Dept. of Biomedical Engineering, University Hospital of Northern Sweden, S-901 85 Umea, Sweden

18. Numerical Instabilities in Bone Remodeling Simulations: The Advantages of a Node-Based Finite Element Approach. Jacobs CR, et al., *J Biomech* 28(4):449–459, 1995.

Contact: Christopher R. Jacobs, Dept. of Orthopaedics, P.O. Box 850, 500 University Dr., Hershey, PA 17033

19. Optimal Control Model for Analyzing Human Postural Balance. Kuo AD, *IEEE Trans Biomed Eng* 42(1):87–101, 1995.

Contact: Arthur D. Kuo, Dept. of Mechanical Engineering and Applied Mechanics, University of Michigan, Ann Arbor, MI 48109-2125

COMMUNICATION AIDS—HEARING

20. Accuracy of Speech Intelligibility Index Predictions for Noise-Masked Young Listeners with Normal Hearing and for Elderly Listeners with Hearing Impairment. Hargus SE, Gordon-Salant S, *J Speech Hear Res* 38(1):234–243, 1995.
Contact: Sandra Gordon-Salant, PhD, University of Maryland at College Park, Dept. of Hearing and Speech Sciences, Lefrak Hall, College Park, MD 20742

21. Communication with Deaf Patients–Knowledge, Beliefs, and Practices of Physicians. Ebert DA, Heckerling PS, JAMA 273(3):227–229, 1995.
Contact: David A. Ebert, MD, Dept. of Medicine, M/C 787, University of Illinois at Chicago, 840 S. Wood St., Chicago, IL 60612

22. Evaluation of a Wide Range of Amplitude-Frequency Responses for the Hearing Impaired. van Buuren RA, Festen JM, Plomp R, J Speech Hear Res 38(1):211–221, 1995.
Contact: Ronald A. van Buuren, Experimental Audiology/ Dept. of Oto-Rhinolaryngology, Free University Hospital, P.O. Box 7057, 1007 MB Amsterdam, The Netherlands

23. Speech Recognition in Amplitude-Modulated Noise of Listeners with Normal and Listeners with Impaired Hearing. Eisenberg LS, Dirks DD, Bell TS, *J Speech Hear Res* 38(1):222–233, 1995.
Contact: Laurie S. Eisenberg, UCLA School of Medicine, Division of Head and Neck Surgery, 31-24 Rehabilitation Ctr., Los Angeles, CA 90024

COMMUNICATION AIDS—SPEECH

24. Errors and Error Correction in Automatic Speech Recognition Systems. Noyes JM, Frankish CR, *Ergonomics* 37(11):1943–1957, 1994.
Contact: J.M. Noyes, Dept. of Psychology, University of Bristol, 8 Woodland Rd., Bristol BS8 1TN, UK

25. Presence, Completeness, and Accuracy of Main Concepts in the Connected Speech of Non-Brain-Damaged Adults and Adults with Aphasia. Nicholas LE, Brookshire RH, *J Speech Hear Res* 38(1):145–156, 1995.
Contact: Linda E. Nicholas, MA, Aphasia Research (127A), VA Medical Ctr., One Veterans Dr., Minneapolis, MN 55417

COMMUNICATION AIDS—VISION

26. Computers, Automation, and the Employment to Persons Who Are Blind or Visually Impaired. Mather J, *J Visual Impairm Blindn* 88(6):544–549, 1994.
Contact: Jacqueline Mather, BA, School of Education, Australian Catholic University, 40 Edward St., North Sydney, 2059 Australia

27. Effectiveness of an Independent Living Services Program for Older Persons Who Are Blind. Farish J, Wen S-S, *J Visual Impairm Blindn* 88(6):525–531, 1994.
Contact: Jean Farish, RhD, CRC, Rehabilitation Counselor Training Program, Special Education and Rehabilitation Services, School of Education, Jackson State University, 1325 Lynch St., Box 17501, Jackson, MS 39217-7501

28. Suppression of Melatonin Secretion in Some Blind Patients by Exposure to Bright Light. Czeisler CA, et al., *N Engl J Med* 332(1):6–11, 1995.
Contact: Charles A. Czeisler, PhD, MD, Section on Sleep Disorders and Circadian Medicine, Division of Endocrinology, Dept. of Medicine, Harvard Medical School and Brigham and Women's Hospital, 221 Longwood Ave., Boston, MA 02115

29. Visual Neuroprosthetics—Functional Vision for the Blind. Normann RA, *IEEE Eng Med Biol Mag* 14(1):77–83, 1995.
Contact: Richard A. Normann, Dept. of Bioengineering, John Moran Laboratories for Applied Vision and Neural Sciences, University of Utah, Salt Lake City, UT 84112

Listings #30 through #109 have been collected for this section by Gregory L. Goodrich, PhD, VA Western Blind Rehabilitation Center, Palo Alto, CA.

A Literature search was conducted using the keyword citation index of Low Vision—The References edited by Gregory L. Goodrich, PhD and Randy T. Jose, OD. Published by the Lighthouse Inc., New York, NY.

Because these listings are extracted from several databases, we lack *Contact* information. If more information is needed, the author may be contacted in care of the publisher of the journal or book in which the article appears.

30. Aetiology of Severe Visual Impairment and Blindness in Microphthalmos. Elder MJ, *Br J Ophthalmol* 78:332–334, 1994.

31. Applying Video and Microcomputer Technology in a Low Vision Setting. Goodrich GL, *Opthalmol Clin North Am* 7(2):177–185, 1994.

32. Assessment of Illumination in Homes of Visually Impaired Individuals. Lau SA, *J Vis Rehabil* 8(4):8–9, 1994.

33. Assessment of Visually Impaired Infants. Hyvarinen L, *Ophthalmol Clin North Am* 7(2):219–225, 1994.

34. Basic Low Vision Examination. Colenbrander A, *Ophthalmol Clin North Am* 7(2):151–162, 1994.

35. Bill of Rights for Patients with Low Vision. Corn AL, *Ophthalmol Clin North Am* 7(2):117–120, 1994.

36. Blindness from Uveitis in a Hospital Population in Sierra Leone. Ronday MJH, et al., *Br J Ophthalmol* 78:690–693, 1994.

37. Causes of Emotional and Psychological Reactions to Adventitious Blindness. Hudson D, *J Vis Impairm Blindn* 88(6):498–503, 1994.

38. Clear Image II Telephoto Microscope. Freeman PB, *J Vis Rehabil* 8(4):4–7, 1994.

39. Cognitive Visual Dysfunction. Dutton GN, *Br J Ophthalmol* 78:723–726, 1994.

40. Comparison of Automated Static Dark Stimuli with the Humphrey STATPAC Program in Glaucomatous Visual Field Loss. Mutlukan E, *Br J Ophthalmol* 78:175–184, 1994.

41. Computers, Automation, and the Employment of Persons Who are Blind or Visually Impaired. Mather J, *J Vis Impairm Blindn* 88(6):544–549, 1994.

42. Concept of Adjustment: A Structural Model. Dodds A, et al., *J Vis Impairm Blindn* 88(6):487–497, 1994.

43. Considerations in Addressing the Daily Living Needs in Older Persons with Low Vision. Warren M, Lampert J, *Ophthalmol Clin North Am* 7(2):187–195, 1994.

44. Contrast Sensitivity Testing in Clinical Practice. Moseley MJ, Hill AR, *Br J Ophthalmol* 78:795–797, 1994.

45. Dark Adaptation in Retinal Abnormalities. Fitzke FW, *Br J Ophthalmol* 78:426, 1994.

46. Deficits in Perimetric Performance in Patients with Symptomatic Human Immunodeficiency Virus Infection or Acquired Immunodeficiency Syndrome. Geier SA, et al., *Am J Ophthalmol* 119(3):335–344, 1995.

47. Design of a Zoom Stand Magnifier–A New Low Vision Device. Spitzberg LA, Chen S, *Optom Vis Sci* 71(10):613–618, 1994.

48. Distribution and Aetiology of Blindness and Visual Impairment in Mesoendemic Onchocercal Communities. Kaduna State, Nigeria. Abiose A, et al., *Br J Ophthalmol* 78:8–13, 1994.

49. Effect of Diabetes Associated Increases in Lens Optical Density on Colour Discrimination in Insulin Dependent Diabetes. Hardy KJ., et al., *Br J Ophthalmol* 78:754–756, 1994.

50. Development of a Scale to Facilitate Reading Mode Decisions. Sharpe M, McNear D, Bosma J, *J Vis Impairm Blindn* 89(1):83–89, 1995.

51. Effectiveness of an Independent Living Services Program for Older Persons Who Are Blind. Farish J, Sen S-S, *J Vis Impairm Blindn* 88(6):525–531, 1994.

52. Emotion and Cognition: Attitudes Toward Persons Who Are Visually Impaired. Verplanken B, Meijnders A, van de Wege A, *J Vis Impairm Blindn* 88(6):504–511, 1994.

53. Evaluation of the Humphrey FASTPAC Threshold Program in Glaucoma. O'Brien C, et al., *Br J Ophthalmol* 78:516–519, 1994.

54. Functional Vision Loss. Hoffman DL, Wilson R, *J Am Optom Assoc* 65(12):835–844, 1994.

55. Future Developments in Computer-Aided Teaching of Visually Impaired Children. Douglas G, Gamble A, *Br J Vis Impairm* 12(3):96–100, 1994.

56. Grating Acuity Tests Should Not be Used for Social Service Purposes in Preliterate Children. Kushner BJ, *Arch Ophthalmol* 112(8):1030–1031, 1994.

57. Housing Design and Persons with Visual Impairment: Report of Focus-Group Discussions. Long RG, *J Vis Impairm Blindn* 89(1):59–69, 1995.

58. Impact of Mother's Interactions on the Development of Their Young Visually Impaired Children. Dote-Kwan J, *J Vis Impairm Blindn* 89(1):46–58, 1995.

59. Information Centers for Newly Blind Persons in Israel. Dickstein N, Gozovsky M, *J Vis Impairm Blindn* 88(6):512–515, 1994.

60. Longitudinal Case Study of Age-Related Macular Degeneration. Gormezano SR, *J Vis Rehabil* 8(4):11–18, 1994.

61. Loss of Vision Before Ophthalmic Referral in Blind and Partially Sighted Diabetics in Bristol. Clark JB, et al., *Br J Ophthalmol* 78:741–744, 1994.

62. Low Vision Aids Training in the Home. Stoll S, Sarma S, Hoeft WW, *J Am Optom Assoc* 66(1):32–38, 1995.

63. Low Vision and Quality of Life: Aspects of Vision Loss. Colenbrander A, *Ophthamol Clin North Am* 7(2):127–130, 1994.

64. Low Vision Magnifiers–Their Optical Parameters and Methods for Prescribing. Bailey IL, et al., *Optom Vis Sci* 71(11):689–698, 1994.

65. Low Vision Rehabilitation in the Elderly. Sterns GK, *Ophthalmol Clin North Am* 7(2):231–236, 1994.

66. Macular Sparing as a Perimetric Artifact. Bischoff P, Lang J, Huber A, *Am J Ophthalmol* 119(1):72–80, 1995.

67. Medications Used by Students with Visual and Multiple Impairments: Implications for Teachers. Kelley P, Wedding JA, *J Vis Impairm Blindn* 89(1):38–45, 1995.

68. New Ergonomic Stand Magnifiers. Spitzberg LA, Goodrich GL, *J Am Optom Assoc* 66(1):25–30, 1995.

69. Nothing More Can Be Done...A Fable for Our Times. Tanton JH, *Ophthalmol Clin North Am* 7(2):203–205, 1994.

70. Nystagmus: Neglected No Longer? Sanders J, *Br J Vis Impairm* 12(3):105–106, 1994.

71. Office Low Vision. Christiansen RM, *Ophthalmol Clin North Am* 7(2):207–209, 1994.

72. Ophthalmologist's Role in the Vision Rehabilitation Process. Ringering LS, *Ophthalmol Clin North Am* 7(2):137–142, 1994.

73. Optometric Contributions to the Vision Rehabilitation Network. Hoeft W, *J Am Optom Assoc* 66(1)14–15, 1995.

74. Optometric Management of Congenital Cataracts. Moore, Bruce D, *J Am Optom Assoc* 65(10):719–724, 1994.

75. Optometrist and the Practice of Low Vision Care. Goodlaw E, *J Am Optom Assoc* 66(1)12–13, 1995.

76. Parents' Role Regarding Their Child with Low Vision. Liegner JT, *Ophthalmol Clin North Am* 7(2):227–230, 1994.

77. Preferred Retinal Locus: A Review with Applications in Vision Rehabilitation. Schuchard RA, Fletcher DC, *Ophthalmol Clin North Am* 7(2):243–256, 1994.

78. Prevalence of Photopsias and Charles Bonnet Syndrome in a Low Vision Population. Crane WG, Fletcher DC, Schuchard RA, *Ophthalmol Clin North Am* 7(2):143–149, 1994.

79. Psychological Ramifications of Losing an Eye. Perez CY, *J Vis Rehabil* 8(4):19–23, 1994.

80. Psychosocial Adjusment to Age-Related Macular Degeneration. Davis C, Lovie-Kitchin J, Thompson B, *J Vis Impairm Blindn* 89(1):16–27, 1995.

81. Quantifying Low Vision: Ranges of Vision Loss. Colenbrander A, *Ophthalmol Clin North Am* 7(2):131–136, 1994.

82. Reading Rates with Artificial Central Scotomata with and without Spatial Remapping Print. Wensyeen JM, Bedell HE, Loshin DS, *Optom Vis Sci* 72(2):100–114, 1995.

83. Registration of Blind and Partially Sighted People (Editorial). Wormald R, Evans J, *Br J Ophthalmol* 78:733–734, 1994.

84. Representation of the Visual Field in the Occipital Striate Cortex. McFadzean R, et al., *Br J Ophthalmol* 78:185–190, 1994.

85. Rewarding Process: A Case Study in Visual Rehabilitation. Fletcher DC, *Ophthalmol Clin North Am* 7(2):121–122, 1994.

86. Role of Eye Pathology in Vision Rehabilitation. Faye E, *Ophthalmol Clin North Am* 7(2):123–126, 1994.

87. Scanning Laser Ophthalmoscope Macular Perimetry and Applications for Low Vision Rehabilitation Clinicians. Fletcher DC, et al., *Ophthalmol Clin North Am* 7(2):257–265, 1994.

88. Scanning Laser Ophthalmoscopic Analysis of the Pattern of Visual Loss in Age-related Geographicatrophy of the Macula. Sunness JS, Bressler NM, Maguire MG, *Am J Ophtholmol* 119(2):143–151, 1995.

89. Screen-Magnification Software for IBM Compatible Computers: An Overview. Uslan MM, Shragai Y, *J Vis Impairm Blindn JVIB News Service* 89(1):19–22, 1995.

90. Spectacles and Magnifiers. Evans LS, *Ophthalmol Clin North Am* 7(2):163–167, 1994.

91. Status of Low Vision and Driving. Goodrich GL, *J Vis Rehabil* 8(4):9–10, 1994.

92. Technique for Determining Glaucomatous Visual field Progression by Using Animation Graphics. Chauhan BC, et al., *Am J Ophthalmol* 118(4):485–491, 1994.

93. Telescopes for the Visually Impaired. Bailey IL, *Ophthamol Clin North Am* 7(2):169–175, 1994.

94. Training Conundrum: Acquiring and Deploying Skills to Meet Needs in Rehabilitation. Lomas GM, *Br J Vis Impairm* 12(3):83–85, 1994.

95. Use of Light Rail or Light Rapid Transit Systems by Individuals with Severe Visual Impairments. Svendsen K, *J Vis Impairm Blindn* 88(1):69–74, 1994.

96. Unrecognised and Unregistered Visual Impairment. Robinson R, et al., *Br J Ophthalmol* 78:736–740, 1994.

97. Using Applied Behavior Analysis Procedures to Change the Behavior of Students With Visual Disabilities: A Research Review. DeMario NC, Crowley EP, *J Vis Impairm Blindn* 88(6):532–543, 1994.

98. Vibrotactile Perception: Perspective Taking by Children Who Are Visually Impaired. Miletic G, *J Vis Impairm Blindn* 88(6):550–563, 1994.

99. Vision and Driving in the Elderly. Owsley C, *Optom Vis Sci* 71(12):727–735, 1994.

100. Vision and Reading. Rubin GS, *Ophthalmol Clin North Am* 7(2):237–241, 1994.

101. Vision Rehabilitation Service of the Massachusetts Eye and Ear Infirmary. Kraut JA, *Ophthamol Clin North Am* 7(2):211–218, 1994.

102. Vision Rehabilitation Therapy for the Bioptic Driver. Politzer MR, *J Am Optom Assoc* 66(1):18–24, 1995.

103. Vision Related Daily Life Problems in Patients Waiting for a Cataract Extraction. Lunstrom M, Fregell G, Sjoblom A, *Br J Ophthalmol* 78:608–611, 1994.

104. Vision Requirements for Driving. Low Vision Committee of the American Academy of Ophthalmology. *Ophthalmol Clin North Am* 7(2):197–201, 1994.

105. Visual Impairment and Disability in Older Adults. Rubin GS, et al., *Optom Vis Sci* 71(12):750–760, 1994.

106. Visual Impairment and Rehabilitation Needs of Nursing Home Residents. Horowitz A, et al., *J Vis Impairm Blindn* 89(1):7–15, 1995.

107. Visual Recovery in a Patient with Leber Hereditary Optic Neuropathy and the 14484 Mutation. Hrynchak PK, Spafford MM, *Optom Vis Sci* 71(10): 604–612, 1994.

108. Visual Rehabilitation in Low Vision Patients with Aging Macular Degeneration. Wu D-Z, et al., *J Am Optom Assoc* 66(1):39–41, 1995.

109. Waterloo Vision and Mobility Study: Gait Adaptations to Altered Surfaces in Individuals with Age-Related Maculopathy. Spaulding SJ, et al., *Optom Vis Sci* 71(12):770–777, 1994.

FUNCTIONAL ASSESSMENT

110. Validity of the Extended Activities of Daily Living Scale. Gompertz P, Pound P, Ebrahim S, *Clin Rehabil* 8(4):275–280, 1994.
Contact: Patrick Gompertz, Dept. of Public Health and Primary Care, Royal Free Hospital School of Medicine, Rowland Hill St., London NW3 2PF, UK

FUNCTIONAL ELECTRICAL STIMULATION

111. Dynamic Performance Model of an Isometric Muscle-Joint Unit. Zhou BH, et al., *Med Eng Phys* 17(2):145–150, 1995.
Contact: Prof. M. Solomonow, Dept. of Orthopaedic Surgery, Louisiana State University Medical Center, 2025 Gravier St., Suite 400, New Orleans, LA 70112

112. Functional Electrical Stimulation. Larsson L-E, *Scand J Rehabil Med 30(Suppl.)*63–72, 1994.
Contact: Lars-Erik Larsson, MD, Dept. of Clinical Neurophysiology, University Hospital, Linkoping, Sweden

GAIT ANALYSIS

113. Biomechanical Model to Determine Lumbosacral Loads During Single Stance Phase in Normal Gait. Khoo BCC, Goh JCH, Bose K, *Med Eng Phys* 17(1):27–35, 1995.
Contact: Dr. James C. H. Goh, Dept. of Orthopaedic Surgery, National University of Singapore, Lower Kent Ridge Rd., Singapore 0511

114. Can Gait Analysis Improve Gait Training in Stroke Patients. Knutsson E, Scand J Rehabil Med 30(Suppl.):73–80, 1994.
Contact: Evert Knutsson, MD, Dept. of Clinical Neurophysiology, Karolinska Hospital, Stockholm, Sweden

115. Characterization of Human Gait by Means of Body Center of Mass Oscillations Derived from Ground Reaction Forces. Crowe A, et al., *IEEE Trans Biomed Eng* 42(3)293–303, 1995.
Contact: Alan Crowe, Dept. of Medical and Physiological Physics, University of Utrecht, The Netherlands

116. Development of a Balance Context Indicator During Gait. Jullian JM, Peruchon E, Rabischong P, *Gait & Post* 2(4):227–234, 1994.
Contact: J.M. Jullian, INSERM Unite 103, Appareil moteur et Handicap, 395 Ave. des Moulins, 34090 Montpellier, France

117. Gait Characteristics as a Function of Age and Gender. Nigg BM, Fisher V, Ronsky JL, *Gait & Post* 2(4):213–220, 1994.
Contact: Benno M. Nigg, Human Performance Laboratory, The University of Calgary, 2500 University Dr. NW, Calgary, Alberta, T2N 1N4 Canada

118. Measured Versus Predicted Vertical Displacement of the Sacrum During Gait as a Tool to Measure Biomechanical Gait Performance. Kerrigan DC, et al., *Am J Phys Med Rehabil* 74(1):3–8, 1995.
Contact: D. Casey Kerrigan, MD, MS, Division of Physical Medicine and Rehabilitation, Harvard Medical School, 125 Nashua St., Boston, MA 02114

119. Modelling and Gait Evaluation of Asymmetrical-Keel Foot Prosthesis. Allard P, et al., *Med Biol Eng Comput* 33(1):2–7, 1995.
Contact: Paul Allard, PhD, PEng, Laboratoire d'Etude du Mouvement, Centre de Recherche Pediatrique, Hopital Sainte-Justine, 3175 Cote Ste-Catherine, Montreal, PQ H3T 1C5, Canada

120. Predicting the Kinematics and Kinetics of Gait Based on the Optimum Trajectory of the Swing Limb. Chou LS, Song SM, Draganich LF, *J Biomech* 28(4):377–385, 1995.
Contact: Louis F. Draganich, PhD, University of Chicago, Section of Orthopaedics and Rehabilitation Medicine, Dept. of Surgery, MC 3079, 5841 South Maryland Ave., Chicago, IL 60637

121. Recovery of Walking Function in Stroke Patients: The Copenhagen Stroke Study. Jorgensen HS, et al., *Arch Phys Med Rehabil* 76(1):27–32, 1995.
Contact: Henrik Stig Jorgensen, MD, Dept. of Neurology, Bispebjerg Hospital, Bispebjerg Bakkle 23, DK-2400 Copenhagen NV, Denmark

122. Simple Device for Measurement of Load-Bearing Forces During Walking. Harrison AJ, Atkins RM, Scott HJF, *Proc Instn Mech Engrs—Part H: J En* 208(H3):159–162, 1994.
Contact: A.J. Harrison, BEng, PhD, Dept. of Mechanical Engineering, University of Bristol, Park Row, Bristol BS1 5LS, UK

123. Three-Dimensional Quantification of Human Standing Posture. Newton RU, Neal RJ, *Gait & Post* 2(4):205–212, 1994.
Contact: Robert U. Newton, Centre for Exercise Science and Sports Management, Southern Cross University, P.O. Box 157, Lismore, NSW, 2480 Australia

See also 3

GENERAL

124. Analytical Model to Predict the Electric Field and Excitation Zones Due to Magnetic Stimulation of

Peripheral Nerves. Ruohonen J, Ravazzani P, Grandori F, *IEEE Trans Biomed Eng* 42(2):158–161, 1995.
Contact: Jarmo Ruohonen, Systems Theory Centre (CNR) and Dept. of Biomedical Engineering, Polytechnic of Milan, I-20133 Milan, Italy

125. Automatic Device to Measure Time Spent Upright. Diggory P, et al., *Clin Rehabil* 8(4):353–357, 1994.
Contact: J. Schwarz, National Research Institute for Gerontology, North West Hospital, Parkville, Melbourne 3052, Australia

126. Computer/GPS Guidance System for Emergency Rescue Vehicle Location. Yang X, Clarke AM, *Biomed Instrum Technol* 29(1):34–38, 1995.
Contact: Alex M. Clarke, PhD, PO Box 70605, Richmond, VA 23255-0605

127. Critical Evaluation of Clinical Research. Katz RT, et al., *Arch Phys Med Rehabil* 76(1):82–93, 1995.
Contact: No Reprints Available

128. Foot Infections in Diabetic Patients: Decision and Cost-Effectiveness Analyses. Eckman MH, et al., *JAMA* 273(9):712–720, 1995.
Contact: Mark H. Eckman, MD, Division of Clinical Decision Making, Dept. of Medicine, New England Medical Center, Box 302, 750 Washington Ave., Boston, MA 02111

129. **Four Methods for Characterizing Disability in the Formation of Function Related Group.** Stineman MG, et al., *Arch Phys Med Rehabil* 75(12):1277–1283, 1994.
Contact: Margaret G. Stineman, MD, Asst. Prof. of Rehabilitation Medicine, University of Pennsylvania, 101 Ralston-Penn Center, 3615 Chestnut St., Philadelphia, PA 19104-2676

130. Future of the Department of Veterans Affairs Health Care Sytem. Fisher ES, Welch HG, *JAMA* 273(8):651–655, 1995.
Contact: Elliot S. Fisher, MD, MPH, VA Outcomes Group (111B), Veterans Affairs Hospital, White River Junction, VT 05009

131. Habitually Wandering Patients. Pankratz L, Jackson J, *N Engl J Med* 331(26):1752–1755, 1994.
Contact: Loren Pankratz, PhD, Psychology Service (116B), VA Medical Center, P.O. Box 1034, Portland, OR 97207

132. Physician Views About Treating Low Back Pain: The Results of a National Survey. Cherkin DC, et al., *Spine* 20(1):1–10, 1995.
Contact: Daniel C. Cherkin, PhD, Center for Health Studies, 1730 Minor Ave., Suite 1600, Seattle, WA 98101

133. Porous-Coated Anatomic (PCA) Total Hip Arthroplasty: A Review of 73 Uncemented Cases with 2-Year Follow Up. Billotti J, et al., *Orthopedics* 18(1):37–43, 1995.
Contact: Mark Zimmerman, PhD, New Jersey Medical School, MSB G578 Orthopedic Surgery, 185 S Orange Ave., Newark, NJ 07103

134. Prevention of Falls Among the Community-Dwelling Elderly: An Overview. Steinmetz HM, Hobson SJG, *Phys Occup Ther Geriatr* 12(4):13–29, 1994.
Contact: Helena M. Steinmetz, BSc(OT), OT(C), Dept. of Occupational Therapy, University of Western Ontario, Elborn College, London, Ontario, N6G 1H1 Canada

135. Process and Outcome of Care: Comparison of Two Inpatient Geriatric Rehabilitation Settings. Bentur N, Eldar R, Davies MA, *Clin Rehabil* 8(4):307–313, 1994.
Contact: Netta Bentur, JDC-Brookdale Institute of Gerontology and Human Development, JDC Hill, POB 13087, Jerusalem 91130, Israel

136. Prolonged Warm Ischemia and Limb Survival: Case Report. Pilcher DB, Seligson D, Davis JH, *J Trauma* 37(6):941–943, 1994.
Contact: David B. Pilcher, MD, 1 South Prospect St., Burlington, VT 05401

137. Psychoremediation of Communication Skills for Cognitively Impaired Persons with Multiple Sclerosis. Foley FW, et al., *J Neurol Rehabil* 8(4):165–176, 1994.
Contact: Frederick W. Foley, PhD, Ferkauf Graduate School of Psychology, Albert Einstein College of Medicine, 1300 Morris Park Ave., Bronx, NY 10461

138. Pulse Oximeter Signal at Various Blood Flow Conditions in an In Vitro Model. Lindberg L-G, et al., *Med Biol Eng Comput* 33(1):87–91, 1995.
Contact: L.-G. Lindberg, Dept. of Biomedical Engineering, University Hospital, S-581 85 Linkoping, Sweden

139. Relationship Between Plantar Pressure Distribution Under the Foot and Insole Comfort. Chen H,

Nigg BM, de Koning J, *Clin Biomech* 9(6):335–341, 1994.
Contact: Benno M. Nigg, Director, Human Performance Laboratory, University of Calgary, 2500 University Dr. NW, Calgary, Alberta T2N 1N4, Canada

140. Shock-Absorbing Effect of Shoe Insert Materials Commonly Used in Management of Lower Extremity Disorders. Shiba N, et al., *Clin Orthop* 310:130–136, 1994.
Contact: Harold B. Kitaoka, MD, Mayo Clinic, 200 First St., SW, Rochester, MN 55905

141. Spectral Electromyographic Assessment of Back Muscles in Patients with Low Back Pain Undergoing Rehabilitation. Roy SH, et al., *Spine* 20(1):38–48, 1995.
Contact: Serge H. Roy, ScD, PT, NeuroMuscular Center, Boston University, Boston, MA 02215

142. Toward a Methodology for Rehabilitation Research. Whyte J, *Am J Phys Med Rehabil* 73(6):428–435, 1994.
Contact: John Whyte, MD, PhD, Moss Rehabilitation Research Institute, 1200 W. Tabor Rd., Philadelphia, PA 19141

143. Why Rehabilitation Research Does Not Work (As Well as We Think it Should). Ottenbacher KJ, *Arch Phys Med Rehabil* 76(2):123–129, 1995.
Contact: Kenneth J. Ottenbacher, PhD, SUNY at Buffalo, Dept. of Rehabilitation Medicine, Center for Functional Assessment Research, 232 Parker Hall, 3435 Main St., Buffalo, NY 14214

GERIATRICS

144. Ambulatory Ability After Hip Fracture: A Prospective Study in Geriatric Patients. Koval KJ, et al., *Clin Orthop* 310:150–159, 1994.
Contact: Kenneth J. Koval, MD, Dept. of Orthopaedic Surgery, Hospital for Joint Diseases Orthopaedic Institute, 301 E. 17th St., New York, NY 10003

145. Biomechanical Assessment of Quiet Standing and Changes Associated with Aging. Panzer VP, Bandinelli S, Hallett M, *Arch Phys Med Rehabil* 76(2):151–157, 1995.
Contact: Mark Hallett, MD, Bldg. 10, Room SN226, NINDS, NIH, Bethesda, MD 20892-1428

146. Cervical Spine Injuries in the Elderly. Lieberman IH, Webb JK, *J Bone Joint Surg* 76B(6):877–881, 1994.
Contact: J.K. Webb, FRCS, Consultant Spinal Surgeon, Centre for Spinal Studies and Surgery, University Hospital, Queen's Medical Centre, Clifton Blvd., Nottingham NG7 2UH, UK

147. Correlation Between Physical Activity and the Gait Characteristics and Ankle Joint Flexibility of the Elderly. Ronsky JL, Nigg BM, Fisher V, *Clin Biomech* 10(1):41–49, 1995.
Contact: Janet Ronsky, Human Performance Laboratory, University of Calgary, 2500 University Dr. NW, Calgary, Alberta, T2N 1N4 Canada

148. Influence of Aging on Leg Muscle Reflex Responses to Stance Perturbation. Nardone A, et al., *Arch Phys Med Rehabil* 76(2):158–165, 1995.
Contact: Antonio Nardone, MD, PhD, Posture and Movement Laboratory, Division of Physical Therapy and Rehabilitation, Clinica de Lavoro Foundation, IRCCS, Medical Center of Rehabilitation, 28010 Veruno (NO), Italy

149. Inter-Rater Agreement and Stability of Functional Assessment in the Community-Based Elderly. Ottenbacher KJ, et al., *Arch Phys Med Rehabil* 75(12):1297–1301, 1994.
Contact: Kenneth J. Ottenbacher, PhD, OTR, SUNY at Buffalo, Center for Functional Assessment Research, 232 Parker Hall, Buffalo, NY 14214

150. Sacral Insufficiency Fractures in the Elderly. Gotis-Graham I, et al., *J Bone Joint Surg* 76B(6):882–886, 1994.
Contact: I. Gotis-Graham, MB BS, Research Scholar of the Arthritis Foundation of Australia, School of Pathology, 4th Fl., Wallace Wurth Bldg., University of New South Wales, Sydney 2052, Australia

See also 10

HEAD TRAUMA AND STROKE

151. Ankle Spasticity is Inversely Correlated with Antagonist Voluntary Contraction in Hemiparetic Subjects. Levin MF, Hui-Chan C, *Electromyogr Clin Neurophysiol* 34(7):415–425, 1994.
Contact: Dr. Mindy F. Levin, Centre de Recherche, Institut de Readaptation de Montreal, 6300 av. Darlington, Montreal, H3S 2J4, Canada

152. Artificial Feedforward as Preparatory Motor Control in Postictal Hemiparesis. Baykousheva-Mateva V, Mandaliev A, *Electromyogr Clin Neurophysiol* 34(7):445–448, 1994.
Contact: Vesselina St. Baykoucheva, MD, Neurologic Clinic, Medical University, V.A. Aprilov Str. 15 A, 4000 Plovdiv, Bulgaria

153. Availability and Reliabiltiy of Information on the Premorbid Functional Status of Stroke Patients in Hospital. Benbow SJ, et al., *Clin Rehabil* 8(4):281–285, 1994.
Contact: S. Benbow, Pain Research Institute, Rice Lane, Liverpool L9 1AE, UK

154. Comparison of Neuropsychological and Functional Outcome, and Uptake of Rehabilitation Services, Following Severe and Moderate Head Injury. Anderson SI, et al., *Clin Rehabil* 8(4):301–306, 1994.
Contact: Prof. J.D. Miller, Dept. of Clinical Neurosciences, Western General Hospital, Crewe Rd. South, Edinburgh EH4 2XU, UK

155. Ipsilateral Upper and Lower Extremity Response Time Among Stroke Patients. Yaretzky A, et al., *Clin Rehabil* 8(4):293–300, 1994.
Contact: Abraham Yaretzky, Geriatric Dept., Sapir Medical Center, 21 Geller St., Kfar Saba, Israel

156. Observing Seated Posture After Stroke: A Reliability Study. Carr EK, Kenney FD, *Clin Rehabil* 8(4):329–333, 1994.
Contact: E.K. Carr, Dept. of Nursing Studies, King's College London, Cornwall House Annexe, Waterloo Rd., London SE1 8TX, UK

157. Potential Reversibility of Hemiparetic Dyskinesias in Adults: Surgical Methods of Amelioration in the Lower Limb. Treanor WJ, *J Neurol Rehabil* 8(4):203–216, 1994.
Contact: Walter J. Treanor, MD, 1370 Spring St., Santa Rosa, CA 95404

MUSCLES, LIGAMENTS, AND TENDONS

158. Adaptation of Skeletal Muscle to Increased Neuromuscular Activity as Induced by Chronic Low Frequency Stimulation. Pette D, *Scand J Rehabil Med* 30(Suppl.)7–18, 1994.
Contact: Dirk Pette, MD, Faculty of Biology, University of Konstanz, Konstanz, Germany

159. Effects of Lateral Trunk Bending on Muscle Recruitments When Resisting Nonsagittally Symmetric Bending Moments. Lavender SA, et al., *Spine* 20(2):184–191, 1995.
Contact: Steve Lavender, PhD, Dept. of Orthopedic Surgery, Rush-Presbyterian-St. Luke's Medical Center, 1653 West Congress Pkwy., Chicago, IL 60612

160. Evaluation of and Methods to Change Muscle Tone. Hagbarth K-E, *Scand J Rehabil Med* 30(Suppl.) 19–32, 1994.
Contact: Karl-Erik Hagbarth, MD, Dept. of Clinical Neurophysiology, University Hospital, Uppsala, Sweden

161. Lumbosacral Movement in the Sit-and-Reach and in Cailliet's Protective-Hamstring Stretch. Liemohn WP, Sharpe GL, Wasserman JF, *Spine* 19(18):2127–2130, 1994.
Contact: Wendell P. Liemonhn, PhD, Dept. of Exercise Science, University of Tennessee, 1914 Andy Holt Ave., Knoxville, TN 37996-2700

162. Muscle-Joint Models Incorporating Activation Dynamics, Moment-Angle, and Moment-Velocity Properties. Shue G, Crago PE, Chizeck HJ, *IEEE Trans Biomed Eng* 42(2):212–223, 1995.
Contact: Guayhaur Shue, Depts. of Biomedical Engineering and Systems Engineering, Case Western Reserve University, Cleveland, OH 44106

163. Role of the Tendon in the Dynamic Performance of Three Different Load-Moving Muscles. Roleveld K, et al., *Ann Biomed Eng* 22(6):682–691, 1994.
Contact: Prof. M. Solomonow, Dept. of Orthopaedics, LSU Medical Center, 2025 Gravier, New Orleans, LA 70112

164. Some Differences of the Electromyographic Erector Spinae Activity Between Normal Subjects and Low Back Pain Patients During the Generation of Isometric Trunk Torque. Alexiev AR, *Electromyogr Clin Neurophysiol* 34(8):495–499, 1994.
Contact: Assen Alexiev, MD, National Center of Physical Therapy and Rehabilitation, Ovcha Kupel Blvd. 2 V, Sofia 1618, Bulgaria

MYOELECTRIC CONTROL

165. Hermite Expansions of Compact Support Waveforms: Applications to Myoelectric Signals. Lo

Conte LR, Merletti R, Sandri GV, *IEEE Trans Biomed Eng* 41(12):1147–1159, 1994.
Contact: Loredana R. Lo Conte, Politecnico di Torino, Torino 10129, Italy

166. Two-Channel Enhancement of a Multifunction Control System. Kuruganti U, Hudgins B, Scott RN, *IEEE Trans Biomed Eng* 42(1):109–115, 1995.
Contact: Usha Kuruganti, Institute of Biomedical Engineering, University of New Brunswick, Fredericton, NB E3B 5A3 Canada

NEUROLOGICAL DISORDERS

167. Burn-Associated Peripheral Polyneuropathy: A Search for Causative Factors. Margherita AJ, et al., *Am J Phys Med Rehabil* 74(1):28–32, 1995.
Contact: Anthony J. Margherita, MD, Rehabilitation Medicine, University of Washington School of Medicine RJ-30, 1959 NE Pacific St., Seattle, WA 98195

168. Topics in Rehabilitation: Guidelines for Industrial Injuries. Mooney V, *Orthopedics* 18(2):101–102, 1995.
Contact: Vert Mooney, MD, UCSD Dept. of Orthopaedics, Ste 300, 4150 Regents Park Row, La Jolla, CA 92037

169. Treatment of Acquired Muscle Spasticity Using Phenol Peripheral Nerve Blocks. Botte MJ, Abrams RA, Bodine-Fowler SC, *Orthopedics* 18(2):151–159, 1995.
Contact: Michael J. Botte, MD, Dept. of Orthopaedics, UCSD Medical Ctr., 200 W. Arbor Dr., 8-894, San Diego, CA 92103

ORTHOPEDICS

170. Adaptive Finite-Element Approach for Analysis of Bone/Prosthesis Interaction. Hubsch PF, et al., *Med Biol Eng Comput* 33(1):33–37, 1995.
Contact: J. Middleton, University College of Swansea, Dept. of Civil Engineering, Swansea, SA2 8PP, Wales, UK

171. Cemented Versus Uncemented Socket in Hip Arthroplasty: A Radiostereometric Study of 60 Randomized Hips Followed for 2 Years. Onsten I, Carlsson AS, *Acta Orthop Scand* 65(5):517–521, 1994.
Contact: Ingemar Onsten, Dept. of Orthopedics, Malmo General Hospital, Lund University, S-214 01 Malmo, Sweden

172. Conservative Options in the Management of Spinal Disorders, Part I—Bed Rest, Mechanical and Energy-Transfer Therapies. Reitman C, Esses SI, *Am J Orthop* 25(2):109–116, 1995.
Contact: Charles Reitman, PT, MD, Baylor College of Medicine, Houston, TX 76703

173. Distinguishing Wear and Creep in Clinically Retrieved Polyethylene Inserts. McDonald MD, Bloebaum RD, *J Biomed Mater Res* 29(1):1–7, 1995.
Contact: Michael D. McDonald, Bone and Joint Research Laboratory (151F), VA Medical Center, 500 Foothill Blvd., Salt Lake City, UT 84148

174. Easy Extraction of a Fractured Hip Screw: A Technical Note. Husted H, Merrild U, *Acta Orthop Scand* 65(6):651–653, 1994.
Contact: Henrik Husted, Dept. of Orthopedics, Sonderborg Hospital, Sydvang 1, DK-6400 Sonderborg, Denmark

175. Increased Metal Release from Cemented Femoral Components Made of Titanium Alloy—19 Hip Prostheses Followed with Radiostereometry (RSA). Karrholm J, et al., *Acta Orthop Scand* 65(6):599–604, 1994.
Contact: Dr. Johan Karrholm, Dept. of Orthopedics, Sahlgren Hospital, S-413 45 Gothenburg, Sweden

176. Mineralization and pH Relationships in Healing Skeletal Defects Grafted with Demineralized Bone Matrix. Chakkalakal DA, et al., *J Biomed Mater Res* 28(12):1439–1443, 1994.
Contact: D.A. Chakkalakal, Dept. of Orthopaedics, College of Medicine, University of Florida, and Bone Research Laboratory, VA Medical Center, Gainesville, FL 32608-1197

177. Peter Principle in Orthopedics. Rubin G, Cohen-Sobel E, Rubin E, *Am J Orthop* 24(1):30–37, 1995.
Contact: Gustav Rubin MD, FACS, New York College of Podiatric Medicine, New York, NY 10003

178. Postphlebitic Syndrome After Hip Arthroplasty—43 Patients Followed at Least 5 Years. McNally MA, et al., *Acta Orthop Scand* 65(6):595–598, 1994.
Contact: Martin A. McNally, The Queen's University of Belfast, Dept. of Orthopedics, Musgrave Park Hospital, Stockman's Lane, Belfast BT9 7JB, Northern Ireland

179. Radiographic Determination of the Anatomic Hip Joint Center: A Cadaver Study. John JF, Fisher PE, *Acta Orthop Scand* 65(5):509–510, 1994.
Contact: Dr. Joachim F. John, Orthopadische Abteilung der Universitatskliniken Freiburg, Hugstetter Str. 55, 79106 Freiburg i.Br., Germany

180. Rheumatoid Arthritis: New Approaches for Its Evaluation and Management. Semble EL, *Arch Phys Med Rehabil* 76(2):190–201, 1995.
Contact: Elliot L. Semble, MD, Dept. of Medicine, Bowman Gray School of Medicine, Medical Center Blvd., Winston-Salem, NC 27157

PROSTHETICS

181. Bioenergetic Comparison of a New Energy-Storing Foot and SACH Foot in Traumatic Below-Knee Vascular Amputations. Casillas J-M, et al., *Arch Phys Med Rehabil* 76(1):39–44, 1995.
Contact: Jean-Marie Casillas, MD, Service de reeducation Fonctionnelle, Centre Hospitalier Universitaire de Dijon, 23, rue Gaffarel, 21034 Dijon, Cedex, France

182. Optimal Control for an Above-Knee Prosthesis with Two Degrees of Freedom. Popovic D, Oguztoreli MN, Stein RB, *J Biomech* 28(1):89–98, 1995.
Contact: Dr. Richard B. Stein, Division of Neuroscience, University of Alberta, 513 Heritage Medical Research Centre, Edmonton, Alberta, T6G 2S2, Canada

183. Treatment of Femoral Fracture Above a Knee Prosthesis—18 Cases Followed 0.5-14 Years. Garnavos C, Rafiq M, Henry APJ, *Acta Orthop Scand* 65(6):610–614, 1994.
Contact: Christos Garnavos, Orthopedic Surgeon, Leighton Hospital, Crewe, Cheshire, CW1 4QJ, UK

SPINAL CORD INJURY

184. Alteration of Immune System Function in Tetraplegics: A Pilot Study. Compagnolo DI, et al., *Am J Phys Med Rehabil* 73(6):387–393, 1994.
Contact: Denise I. Campagnolo, MD, Dept. of Physical Medicine and Rehabilitation, University Hospital, B261, 150 Bergen St., Newark, NJ 07103-2406

185. Assessment of Adaptive Transportation Technology: A Survey of Users and Equipment Vendors. Sprigle S, et al., *Assist Technol* 6(2):111–119, 1994.
Contact: Stephen Sprigle, PhD, Center for Assistive Technology, 515 Kimball Tower, SUNY-Buffalo, Buffalo, NY 14214

186. Diaphyseal Femoral Shortening for the Treatment of Hip and Knee Extension in a Spastic Quadriplegic. Tedder JL, Insler HP, Adar U, *Orthopedics* 17(12):1141–1143, 1994.
Contact: Jeffrey L. Tedder, MD, 1114 Roseate Ct., Bradenton, FL 34209

187. Effect of Body Training Position on Outcomes of an Aerobic Training Study on Individuals with Quadriplegia. McLean KP, Skinner JS, *Arch Phys Med Rehabil* 76(2):139–150, 1995.
Contact: Reprints Not Available

188. New Approach to Reaching Control for Tetraplegic Subjects. Popovic M, Popovic D, *J Electromyograph Kinesiol* 4(4):242–253, 1994.
Contact: Mrs. Mirjana Popovic, The Miami Project, University of Miami School of Medicine, 1600 NW 10th Ave., R-48, Miami, FL 33136

189. Possibilities to Evaluate and Diminish the Effects of the Trauma in Spinal Cord Lesions: An Experimental Study in the Rat. Winkler T, *Scand J Rehabil Med* 30(Suppl.):81–82, 1994.
Contact: Tomas Winkler, MD, Dept. of Clinical Neurophysiology, University Hospital, S-751 85 Uppsala, Sweden

190. Prevalence and Impact of Wrist and Shoulder Pain in Patients with Spinal Cord Injury. Subbarao JV, Klopfstein J, Turpin R, *J Spinal Cord Med* 18(1):9–13, 1994.
Contact: Jay V. Subbarao, MD, MS, Director, Comprehensive Rehabilitative Services (11R), Edward Hines Jr. VA Hospital, Hines, IL 60141

191. Properties of the Venous Vascular System in the Lower Extremities of Individuals with Paraplegia. Hopman MTE, et al., *Paraplegia* 32(12):810–816, 1994.
Contact: M.T.E. Hopman, MD, PhD, Asst. Prof. of Physiology, University of Nijmegen, P.O. Box 9101, 6500 HB Nijmegen, The Netherlands

192. Use of the Audiospinal Response to Test for Completeness of Spinal Cord Injury. Dobkin BH, Taly AB, Su G, *J Neurol Rehabil* 8(4):187–191, 1994.
Contact: Bruce H. Dobkin, MD, Dept. of Neurology, School of Medicine, University of California, Los Angeles, 300 UCLA Medical Plaza, Los Angeles, CA 90024-6975

SURGERY

193. Results After Replantation and Revascularization in the Upper Extremity in Children. Saies AD, et al., *J Bone Joint Surg* 76A(12):1766–1776, 1994.
Contact: A.D. Saies, MBBS, FRACS, Sportsmed-SA, 32 Payneham Rd., Stepney 5069, South Australia, Australia

See also 4

VASCULAR DISORDERS

194. Empirical Technique to Compensate for Melanin When Monitoring Skin Microcirculation Using Reflectance Spectrophotometry. Ferguson-Pell M, Hagisawa S, *Med Eng Phys* 17(2):104–110, 1995.
Contact: Martin Ferguson-Pell, PhD, Helen Hayes Hospital, Center for Rehabilitation Technology, Route 9W, West Haverstraw, NY 10993

195. Standing Balance in Trans-Tibial Amputees Following Vascular Disease or Trauma: A Comparative Study with Healthy Subjects. Hermodsson Y, et al., *Prosthet Orthot Int* 18(3):150–158, 1994.
Contact: Y. Hermodsson, RPT, Dept. of Orthopaedics, Helsingborg Hospital, S-251 87 Helsingborg, Sweden

WHEELCHAIRS and POWERED VEHICLES

196. Kinematics of Wheelchair Propulsion in Adults and Children with Spinal Cord Injury. Bednarczyk JH, Sanderson DJ, *Arch Phys Med Rehabil* 75(12):1327–1334, 1994.
Contact: David J. Sanderson, PhD, Biomechanics Laboratory, School of Human Kinetics, University of British Columbia, 210-6081 University Blvd., Vancouver, BC V6T 1Z1 Canada

197. Lifestyle Implications of Power Mobility. Miles-Tapping C, MacDonald LJ, *Phys Occup Ther Geriatr* 12(4):31–49, 1994.
Contact: Carol Miles-Tapping, PhD, OTM, O.T. Dept., Rehabilitation Services, Victoria General Hospital, 2340 Pembina Highway, Winnipeg, Manitoba, R3E 2E8 Canada

198. Positioning and Securement of Riders and Their Mobility Aids in Transit Vehicles: An Analytical Review. Karg P, Yaffe K, Berkowitz D, *Assist Technol* 6(2):94–110, 1994.
Contact: Patricia Karg, MSBME, Project Engineer, Health Devices System, ECRI, 5200 Butler Pike, Plymouth Meeting, PA 19462

WOUNDS and ULCERS

199. Management of Traumatic Cutaneous Defects by Using a Skin-Stretching Device. Boden BP, Buinewicz BR, *Am J Orthop (Suppl.)*:27–30, 1994.
Contact: Barry P. Boden, MD, Resident in Orthopedics, Temple University Hospital, Philadelphia, PA 19122

200. Probing to Bone in Infected Pedal Ulcers: A Clinical Sign of Underlying Osteomyelitis in Diabetic Patients. Grayson ML, et al., *JAMA* 273(9):721–723, 1995.
Contact: Adolf W. Karchmer, MD, Division of Infectious Diseases, New England Deaconess, Hospital, 185 Pilgrim Rd., Boston, MA 02215

Periodicals reviewed for PUBLICATIONS OF INTEREST

Acta Orthopaedica Scandinavica
Advances in Orthopaedic Surgery
American Journal of Occupational Therapy
American Journal of Orthopedics
American Journal of Physical Medicine and Rehabilitation
American Journal of Sports Medicine
American Rehabilitation
Annals of Biomedical Engineering
Archives of Physical Medicine and Rehabilitation
ASHA (American Speech and Hearing Association)
Assistive Technology
Biomaterials, Artificial Cells, and Immobilization Biotechnology
Biomedical Instrumentation & Technology

Canadian Journal of Rehabilitation
Clinical Biomechanics
Clinical Kinesiology
Clinical Orthopaedics and Related Research
Clinical Rehabilitation
CRC Critical Reviews in Biomedical Engineering
DAV Magazine (Disabled American Veterans)
Disability and Rehabilitation
Electromyography and Clinical Neurophysiology
Engineering in Medicine and Biology Magazine
Ergonomics
Gait and Posture
Hearing Journal
Hearing Rehabilitation Quarterly
Hearing Research
Human Factors: The Journal of the Human Factors Society
IEEE Engineering in Medicine and Biology Magazine
IEEE Transactions on Biomedical Engineering
IEEE Transactions in Systems, Man and Cybernetics
IEEE Transactions on Rehabilitatin Engineer
International Journal of Rehabilitation Research
JAMA
Journal of Acoustical Society of America
Journal of Applied Biomaterials
Journal of Biomechanical Engineering
Journal of Biomechanics
Journal of Biomedical Engineering
Journal of Biomedical Materials Research
Journal of Bone and Joint Surgery—American Ed.
Journal of Bone and Joint Surgery—British Ed.
Journal of Clinical Engineering
Journal of Electromyography and Kinesiology
Journal of Head Trauma and Rehabilitation
Journal of Medical Engineering and Technology
Journal of Neurologic Rehabilitation
Journal of Orthopaedic and Sports Physical Therapy
Journal of Orthopaedic Research
Journal of Prosthetics and Orthotics
Journal of Rehabilitation
Journal of Speech and Hearing Research
Journal of Spinal Cord Medicine
Journal of Trauma
Journal of Vision Rehabilitation
Journal of Visual Impairment and Blindness
The Lancet
Medical and Biological Engineering and Computing
Medical Engineering Physics
Medical Psychotherapy Yearbook
Medicine & Science in Sports and Exercise
Military Medicine
New England Journal of Medicine
The Occupational Therapy Journal of Research
Orthopaedic Review
Orthopedic Clinics of North America
Orthopedics
Paraplegia
Paraplegia News
Physical and Occupational Therapy in Geriatrics
Physical Medicine and Rehabilitation
Physical Therapy
Physiotherapy
Proceedings of the Institution of Mechanical Engineers—Part H: Journal of Engineering in Medicine
Prosthetics and Orthotics International
Rehab Management
Rehabilitation Digest
Scandinavian Journal of Rehabilitation Medicine
Science
Spine
Sports 'N Spokes
Techniques in Orthopaedics
Topics in Geriatric Rehabilitation
VA Practitioner
Vanguard
Volta Review

CALENDAR OF EVENTS

NOTE: An asterisk at the end of a citation indicates a new entry to the calendar.

1995

May 27–31, 1995
1st International Rehabilitation Medicine Congress, Istanbul, Turkey
Contact: Dr. Onder Kayhan, Congress Secretariat, P.O. Box 1, Kosuyolu 81121, Istanbul, Turkey; Tel: (90) (216)326–4217; Fax: (90) (216)325–0323

May 28–31, 1995
5th National Home Care Conference, Edmonton, Alberta
Contact: Conference Secretariat, c/o The CanaGlobe Group Inc., 15016 77 Avenue, Edmonton, AB, T5R 3B3; Tel: 403-487-8102; Fax: 403-487-2417*

May 28–June 1, 1995
5th European Congress on Research in Rehabilitation, Helsinki, Finland
Contact: Professor Simon Miller, Division of Clinical Neuroscience, The Medical School, The University, Newcastle upon Tyne, NE2 4HH, United Kingdom; Tel: +44 91 222 6617; Fax: +44 91 222 8803

June 4–7, 1995
11th Annual Meeting, International Society of Technology Assessment in Health Care, Stockholm, Sweden
Contact: ISTAHC Congrex (Sweden) AB, PO Box 5619, S-114 86, Stockholm, Sweden

June 9–14, 1995
RESNA International Conference Vancouver, BC
Contact: RESNA; Tel: 703-524-6686

June 13–17, 1995
15th National Veterans Wheelchair Games, Atlanta, Georgia
Contact: PVA (800) 424-8200, ext. 752 or 687, or VA National Recreation Therapy Events (410) 605-7277

June 23–25, 1995
Annual Meeting of the American Congress of Rehabilitation Medicine, Arlington, Virginia
Contact: American Congress of Rehabilitation Medicine, 5700 Old Orchard Road, First Floor, Skokie, IL 60077-1057; Tel: 708-966-0095; Fax: 708-966-9418

June 25–30, 1995
American Physical Therapy Association, World Congress, Washington, DC
Contact: APTA, 1111 N. Fairfax Street, Alexandria, Virginia 22314-3492; Tel: (800)999-2782*

July 2–3, 1995
5th Conference European Orthopaedic Research Society, Munich, Germany
Contact: Dr. HP Scharf, Orthopaedische Klinik/RKU, Oberer Eselsberg 45, D-89081 Ulm, Germany; Tel: 731 177 513; Fax: 731 177 574

July 2–6, 1995
XVth Congress of the International Society of Biomechanics, Jyvaskyla, Finland
Contact: XVth ISB Congress Secretariat, University of Jyvaskyla, Jyvaskyla Congresses, PO Box 35, Fin-40351 Jyvaskyla, Finland; INTERNET: tvanttin@JYU.FI; FAX: +358 41 603621

July 4–7, 1995
Second Congress of the European Federation of National Associations of Orthopaedics and Traumatology, Munich, Germany
Contact: Administrative Secretariat, Intercongress GmbH, Krenzberger Ring, 66 D-65205 Wiesbaden, Germany; Tel: 49 611 97 87 120; Fax: 49 611 97 87 111*

July 9–16, 1995
4th World Congress of Neuroscience, Kyoto, Japan
Contact: Host Organizer, Secretariat, 4th World Congress of Neuroscience, c/o International Communications, Inc., Kasho Bldg., 2-14-9, Nihonbashi, Chuo-ku, Tokyo 103, Japan; Tel: 03-3272-7981; Fax: 03-3273-2445

July 16–18, 1995
18th International Congress on Education of the Deaf, Jerusalem, Israel
Contact: Secretariat, 18th International Congress on Education of the Deaf/1995, PO Box 50006, Tel Aviv 61500, Israel

July 16–19, 1995
7th International Conference on Mobility and Transport for Elderly and Disabled People, Reading, England
Contact: 7th International Conference Secretariat, Dis-

ability Unit, Department of Transport, Room S10/21, 2 Marsham Street, London SW1P 3EB, England

July 22–27, 1995
3rd International Neurotrauma Symposium, Toronto, Canada
Contact: Conference Secretariat c/o: Congress Canada, 191 Niagara Street, Toronto, Ontario, Canada, M5V 1C9; Tel: 416-860-1772; Fax: 416-860-0380

August 17–19, 1995
5th Vienna International Workshop on Functional Electrostimulation: Basics, Technology and Application, Vienna, Austria
Contact: Institute of Biomedical Engineering and Physics, Secretary: Ch. Jancik, Wahringer Gurtel 18-20/4L, A-1090 Vienna, Austria; Tel: (+43-1)40400/1984; Fax: (+43-1)40400/3988; E-mail: M.BIJAK@BMTP.AKH-WIEN.AC.AT

September 5–7, 1995
41st Annual Conference of the American Paraplegia Society, Las Vegas, Nevada
Contact: Mario T. Balmaseda, Jr., MD, Program Committee Chairman, American Paraplegia Society, 75-20 Astoria Boulevard, Jackson Heights, NY 11370-1177

September 5–8, 1995
2nd Leeds European Rehabilitation Conference Neurological Rehabilitation: New Initiatives in Treatment & Measuring Outcome, Leeds, England UK
Contact: Mrs. Carol Would, Conference Secretary, Department of Continuing Professional Education, Continuing Education Building, Springfield Mount, Leeds LS2 9NG; Tel: (0532) 333232; Fax: (0532) 333240

September 8–10, 1995
4th Scientific Meeting of the Scandinavian Medical Society of Paraplegia, Oslo, Norway
Contact: Congress Secretariat, 4th Scientific of SMSOP, c/o Sunnaas Hospital, N-1450 Nesoddtangen, Norway; Tel: +47 66 96 90 00; Fax: +47 66 91 25 76

September 11–13, 1995
First Biennial Conference, Advancing Human Communication: An Interdisciplinary Forum on Hearing Aid Research and Development, Bethesda, Maryland
Contact: NIDCD; Tel: 301-496-7243; TDD 301-402-0252

September 11–19, 1995
10th Asia Pacific Regional Conference of Rehabilitation International, Indonesia
Contact: Secretariat, 10th ASPARERI, H.Hang, jebat II-2 Blok F IV, Kebayoran Baru, Jakarta 12120, Indonesia; Tel: +62 21 717 366

September 19–23, 1995
American Academy of Orthotists and Prosthetists, National Assembly (AAOP), San Antonio, Texas
Contact: Annette Suriani, 1650 King Street, Suite 500, Alexandria, VA 22314; Tel: 703-836-7116

September 20–23, 1995
17th Annual International Conference of the IEEE Engineering in Medicine and Biology Society and 21st Canadian Medical and Biological Engineering Conference, Montreal, Canada
Contact: Robert E. Kearney, PhD, Eng, Department of Biomedical Engineering, McGill University, 3775 University Street, Montreal, Quebec, Canada H3A 2B4; Tel:514-398-6737; Fax: 514-398-7461; E-Mail: rob@NEURON.BIOMED.MCGILL.CA

October 5–8, 1995
International Conference on Aging and Physical Activity: Promoting Vitality and Wellness in Later Years, Colorado Springs, Colorado
Contact: Laura Wilhelm, Human Kinetics, PO Box 5076, Champaign, IL 61825-5076; Tel: (800) 747-4457 or (217) 351-5076; Fax: (217) 351-2674*

October 9–13, 1995
39th Annual Meeting, Human Factors and Ergonomics Society, San Diego, California
Contact: HFES, PO Box 1369, Santa Monica, CA 90406-1369; Tel: 310-394-1811; Fax: 310-394-2410; E-Mail: 72133.1474@COMPUSERVE.COM*

November 2–4, 1995
Annual Scientific Meeting of the International Medical Society of Paraplegia, New Delhi, India
Contact: The Secretariat, International Medical Society of Paraplegia, National Spinal Injuries Centre, Stoke Mandeville Hospital, Aylesbury, Bucks HP21 8AL, UK; Tel: 44 296 315866; Fax: 44 296 315286

November 12–16, 1995
11th Congress of the Western Pacific Orthopaedic Association, Hong Kong

Contact: Professor SP Chow, Department of Orthopaedic Surgery, Queen Mary Hospital, Pokfulam, Hong Kong; Tel: 852 855 4258; Fax: 852 817 4392

November 15–18, 1995
National Home Health Expo, Atlanta, Georgia
Contact: Home Health Expo; Tel: 404-641-8181*

November 16–20, 1995
American Academy of Physical Medicine & Rehabilitation, Orlando, Florida
Contact: Joan Cahill, AAPM&R, 122 S. Michigan Avenue, Suite 1300, Chicago, IL 60603; Tel: 312-922-9366; Fax: 312-922-6754*

November 17–20, 1995
American Speech-Language-Hearing Association (ASHA), Annual Convention, Cincinnati, Ohio
Contact: Frances Johnston, ASHA, 10801 Rockville Pike, Rockville, MD 20852; Tel: (301) 897-5700

November 30–December 2, 1995
National Conference, The Association for Persons With Severe Handicaps (TASH), San Francisco, California
Contact: Tel: 410-828-TASH*

1996

February 22–27, 1996
American Academy of Orthopedic Surgeons Annual Convention (AAOS), Orlando, Florida
Contact: AAOS, 6300 North River Road, Rosemont, Il 60018-4226; Tel: 708-823-7186; Fax: 708-823-8031

April 22–May 5, 1996
18th World Congress of Rehabilitation International Equality Through Participation—2000 and Beyond, Auckland, New Zealand
Contact: Mrs. Bice Awan, Accident Rehabilitation & Compensation, Insurance Corporation, PO Box 242, Wellington, New Zealand; Tel: +64 4 4738 775

May 12–16, 1996
The First Mediterranean Congress on Physical Medicine and Rehabilitation, Herzlia, Israel
Contact: Dr. Haim Ring, c/o Ortra Ltd., PO Box 50432, Tel Aviv 61500, Israel; Tel: 972-3-664825; Fax: 972-3-660952

May 18–23, 1996
World Congress on Osteoporosis, Amsterdam, The Netherlands
Contact: Congress Secretariat: CONGREX Holland bv, Keizersgracht 782, 1017 EC Amsterdam, NL; Tel: +31.20.6261372; Fax: +31.20.6259574*

August 7–10, 1996
7th International ISAAC Conference on Augmentative and Alternative Communication, Vancouver, BC, Canada
Contact: ISAAC, PO Box 1762, Station R., Toronto, Ontario, Canada; Tel: +1 416 737 9308

August 16–27, 1996
1996 Atlanta Paralympic Games, Atlanta, Georgia
Contact: Tel: 404-588-1996

1997

February 13–18, 1997
American Academy of Orthopedic Surgeons Annual Convention (AAOS), San Francisco, California
Contact: AAOS, 1650 King Street, Suite 500, Alexandria, VA 22314; Tel: 703-836-7114

August 31–September 5, 1997
8th World Congress of the International Rehabilitation Medicine Association IRMA, Kyoto, Japan
Contact: Japan Convention Services, Inc., Nippon Press Center Bldg., 2-1, 2-chome, Uchisaiwai-cho, Chiyoda-ku, Tokyo 100, Japan

*U.S. G.P.O.:1995-387-881:00003

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